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☐ 1. Document ID: US 6194899 B1

L7: Entry 1 of 2 File: USPT Feb 27, 2001

US-PAT-NO: 6194899
DOCUMENT-IDENTIFIER: US 6194899 B1
TITLE: Temperature monitoring method, temperature monitoring apparatus and magnetic resonance apparatus

Full | Title | Citation | Front | Review | Classification | Date | Reference | Claims | KWIC | Draw Desc | Image

☐ 2. Document ID: US 5916161 A

L7: Entry 2 of 2 File: USPT Jun 29, 1999

US-PAT-NO: 5916161
DOCUMENT-IDENTIFIER: US 5916161 A
TITLE: Magnetic resonance imaging apparatus with temperature measurement function

Full | Title | Citation | Front | Review | Classification | Date | Reference | Claims | KWIC | Draw Desc | Image

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Term	Documents
DETECT$	0
DETECT.USPT.	273045
DETECTA.USPT.	3
DETECTAB.USPT.	1
DETECTABALE.USPT.	5
DETECTABE.USPT.	3
DETECTABEL.USPT.	2
DETECTABIE.USPT.	5
DETECTABILILTY.USPT.	1
DETECTABILITIES.USPT.	9
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☐ 1. Document ID: US 6194899 B1

L9: Entry 1 of 4 File: USPT Feb 27, 2001

US-PAT-NO: 6194899
DOCUMENT-IDENTIFIER: US 6194899 B1

TITLE: Temperature monitoring method, temperature monitoring apparatus and magnetic resonance apparatus

DATE-ISSUED: February 27, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Ishihara; Yasutoshi	Tokyo	N/A	N/A	JPX
Umeda; Masaaki	Kawasaki	N/A	N/A	JPX
Watanabe; Hidehiro	Yokohama	N/A	N/A	JPX
Okamoto; Kazuya	Yono	N/A	N/A	JPX

US-CL-CURRENT: 324/315; 600/412

Full | Title | Citation | Front | Review | Classification | Date | Reference | KWIC | Draw. Desc | Image

☐ 2. Document ID: US 6054855 A

L9: Entry 2 of 4 File: USPT Apr 25, 2000

US-PAT-NO: 6054855
DOCUMENT-IDENTIFIER: US 6054855 A

TITLE: Magnetic susceptibility control of superconducting materials in nuclear magnetic resonance (NMR) probes

DATE-ISSUED: April 25, 2000

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Anderson; Weston	Palo Alto	CA	N/A	N/A

US-CL-CURRENT: 324/318; 324/307

Full | Title | Citation | Front | Review | Classification | Date | Reference | KWIC | Draw. Desc | Image

☐ 3. Document ID: US 5378987 A

L9: Entry 3 of 4 File: USPT Jan 3, 1995

US-PAT-NO: 5378987
DOCUMENT-IDENTIFIER: US [redacted]8987 A

TITLE: Method and apparatus for non-invasive measurement of temperature distribution within target body using nuclear magnetic resonance imaging

DATE-ISSUED: January 3, 1995

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Ishihara; Yasutoshi	Kanagawa	N/A	N/A	JPX
Sato; Kozo	Kanagawa	N/A	N/A	JPX

US-CL-CURRENT: 324/315; 600/412

Full | Title | Citation | Front | Review | Classification | Date | Reference | KWIC | Draw Desc | Image

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☐ 4. Document ID: US 4689563 A

L9: Entry 4 of 4 File: USPT Aug 25, 1987

US-PAT-NO: 4689563
DOCUMENT-IDENTIFIER: US 4689563 A

TITLE: High-field nuclear magnetic resonance imaging/spectroscopy system

DATE-ISSUED: August 25, 1987

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Bottomley; Paul A.	Clifton Park	NY	N/A	N/A
Edelstein; William A.	Schenectady	NY	N/A	N/A
Hart, Jr.; Howard R.	Schenectady	NY	N/A	N/A
Schenck; John F.	Schenectady	NY	N/A	N/A
Redington; Rowland W.	Schenectady	NY	N/A	N/A
Leue; William M.	Albany	NY	N/A	N/A

US-CL-CURRENT: 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | KWIC | Draw Desc | Image

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Term	Documents
SHIM.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	20113
SHIMS.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	12450
COIL.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	683202
COILS.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	217562
(8 AND (SHIM WITH COIL)).USPT,PGPB,JPAB,EPAB,DWPI,TDBD.	4

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1. Document ID: US 6194899 B1

L11: Entry 1 of 1 File: USPT Feb 27, 2001

US-PAT-NO: 6194899
DOCUMENT-IDENTIFIER: US 6194899 B1

TITLE: Temperature monitoring method, temperature monitoring apparatus and magnetic resonance apparatus

DATE-ISSUED: February 27, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Ishihara; Yasutoshi	Tokyo	N/A	N/A	JPX
Umeda; Masaaki	Kawasaki	N/A	N/A	JPX
Watanabe; Hidehiro	Yokohama	N/A	N/A	JPX
Okamoto; Kazuya	Yono	N/A	N/A	JPX

US-CL-CURRENT: 324/315; 600/412

Image

Term	Documents
"HIGH FREQUENCY".DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	0
HIGH-FREQUENCY.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	76411
HIGH-FREQUENCIES.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	234
HIGH-FREQUENCYS	0
HF.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	95617
HFS.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	305
(10 AND (HIGH-FREQUENCY OR HF OR "HIGH FREQUENCY")).USPT,PGPB,JPAB,EPAB,DWPI,TDBD.	1

Documents, starting with Document:

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L11: Entry 1 of 1 File: USPT Feb 27, 2001

DOCUMENT-IDENTIFIER: US 6194899 B1
TITLE: Temperature monitoring method, temperature monitoring apparatus and magnetic resonance apparatus

ABPL:
An absolute temperature measuring pulse sequence is executed and, subsequently, a relative temperature measuring pulse sequence is repeatedly executed. Since while a relative temperature can be measured from phase information, an absolute temperature requires frequency information, a time required in the relative temperature measuring pulse can be made shorter than that required in the absolute temperature measuring pulse sequence. Since the relative temperature reveals a temperature variation, if an absolute temperature at a given time is known, an absolute temperature at a subsequent time can be calculated from the relative temperature. Thus, a local internal temperature of the subject can be measured, with a shorter temporal resolution, with the use of the absolute temperature and relative temperature.

BSPR:
The present invention relates to a temperature monitoring method and apparatus which, in order to monitor the internal temperature of a subject during a period of a brain hypothermia primarily applied to a brain disease and cerebropathia, acquire a temperature variation in the inside of the subject by utilizing a magnetic resonance phenomenon and display it and to a magnetic resonance apparatus.

BSPR:
The results of treatment on the brain disorders, such as the brain trauma, brain blood vessel disorder and hypoxia cerebropathia at a time of the stoppage of a cardiopulmonary function are governed by conditions of the brain edema and intracranial pressure exacerbation involved due to the ischemia. For this reason, it is very important to measure the intracranial pressure level and internal jugular sinus venous blood oxygen saturation level and, by doing so, control the subject.

BSPR:
In recent times it has been clarified that, due to an intrabrain heat retention resulting from re-perfusion following the brain disorder a secondary phase of disease appears in advance of the brain edema. It has been, therefore, indicated that it is important to continue treatment while controlling the temperature of the brain.

BSPR:
Attention has recently been paid to the brain hypothermia which, in order to protect to a brain from a brain injury region and resuscitate the brain, lowers the temperature of the brain to 32.degree. to 34.degree. C. and maintains this state for an about 2-days to one-week period (an about two-days to two-weeks period though depending upon the case)--Intensive & critical care medicine Vol. 9, No. 6, 613-689 (1997) issued by Sohgo Igaku Co., Ltd.

BSPR:
It has been difficult to exactly measure the temperature of the brain in real time during the brain hypothermia period. It has been conventional practice to measure the blood temperature primarily responsible for the brain temperature formation, with the use of a catheter sensor set in the internal jugular vein and use it in place of the brain temperature.

BSPR:
It has been reported that, for a normal subject, such internal jugular vein temperature is approximately equal to the brain temperature (Hayashi, Intensive & critical care medicine vol. 7, No. 3, 267-274 (1997)). It has also been reported that a temperature profile is produced in the brain in the case of a patient with the brain disorder. And a non-invasive and high accurate temperature measuring method has thus been desired.

BSPR:
It is accordingly the object of the present invention to measure a local internal temperature in a subject with a short temporal resolution with the use of an absolute temperature.

BSPR:
In the present invention, a pulse sequence for absolute temperature measurement is executed and then a pulse sequence for relative temperature measurement is repeatedly executed. Since the absolute temperature requires frequency information while the relative temperature can be measured from phase information, a time necessary to the pulse sequence for relative temperature measurement can be made shorter than that necessary to the pulse sequence for absolute temperature measurement. Further, the relative temperature reveals a temperature variation and, if an absolute temperature at a given time is found, then that absolute temperature at a sequential time can be calculated from the relative temperature. It is, therefore, possible to measure a local internal temperature in the subject, with a shorter temporal resolution, with the use of the absolute temperature.

BSPR:
Additional objects and advantages of the invention will be set forth in the description which follows, and in part will be obvious from the description, or may be learned by practice of the invention. The objects and advantages of the invention may be realized and obtained by means of the instrumentalities and combinations particularly pointed out hereinafter.

DRPR:
FIG. 1 is a view showing a temperature monitoring apparatus (magnetic resonance apparatus) according to a preferred embodiment of the present invention;

DRPR:
FIG. 2 is a view showing a whole flow of internal temperature monitoring made according to the present embodiment;

DRPR:
FIG. 3A is a view showing an actual positional relation between a temperature monitoring region designated by a step (b) in FIG. 2 and a cerebropathia region;

DRPR:
FIG. 3B is a view showing a positional relation, on an MRI image, between the temperature monitoring region designated by a step (b) in FIG. 2 and the cerebropathia region;

DRPR:
FIG. 4A is a view showing an absolute temperature measuring pulse sequence executed at a step (c) in FIG. 2;

DRPR:
FIG. 5 is a view showing a relative temperature measuring pulse sequence executed at a step (f) in FIG. 2;

DRPR:
FIG. 6A is a view showing a flow of a temperature monitoring pulse sequence executed according to the present embodiment;

DRPR:
FIG. 6B is a view showing a flow of a pulse sequence for a motion error detection function-equipped temperature monitoring feature according to the present embodiment;

DRPR:
FIG. 7A is a view showing a proton spectrum analyzed from an MR signal acquired at an absolute temperature measuring pulse sequence at time t.sub.0 in FIG. 6B; and

DRPR:
FIG. 7B is a view showing a proton spectrum analyzed from an MR signal acquired at an absolute temperature measuring pulse sequence at time t.sub.1 in FIG. 6B.

DEPR:
A temperature monitoring apparatus, or a magnetic resonance apparatus having a temperature monitoring function, according to the present invention will be described in more detail below with reference to the drawing.

DEPR:
FIG. 1 is a block diagram showing an arrangement of the temperature monitoring

apparatus. The internal temperature measurement of the present embodiment is conducted by utilizing the property according to which the resonance frequency of a proton in H.sub.2 O reveals a temperature dependence, that is, varies with the temperature, and the property according to which the resonance frequency of a proton in fat exhibits no temperature dependence, that is, undergoes almost no variation irrespective of the temperature. Needless to say, a target nucleus is not restricted to the proton and to an H.sub.2 O/fat combination one of which a molecule containing such a target nucleus is composed.

DEPR:
A static magnet 1 generates a static magnetic field. Inside the static magnet 1, a shim coil 4, gradient coil 2 and RF probe 3 are arranged.

DEPR:
The shim coil 4 is driven by a shim coil power supply 6 to generate a magnetic field for correcting an inhomogeneous static field. The magnetic field generated from the shim coil 4 is added to the static field and, by doing so, it is possible to enhance the homogeneity of the static magnetic field. It is to be noted that the axis in a coordinate corresponding to the direction of the static field is defined as a z-axis.

DEPR:
The gradient coil 2, being driven by a gradient coil power supply, generates a gradient field on the x-axis, a gradient field on an y-axis and a gradient field on a z-axis. The RF probe 3 is arranged inside the gradient coil 2 and comprises an RF coil and a tuner to allow the target nucleus, here, a proton, to be tuned to a resonance frequency. The RF probe 3 is connected by a duplexer to a transmitter 7 at a transmit time and to a receiver 7 at a receive time.

DEPR:
The transmitter 7 supplies an RF current pulse to the RF probe 3. By doing so, the RF coil generates an RF magnetic field (rotating magnetic field). Further, the receiver 8 receives an MR signal via the RF coil which is generated from a proton spin in the subject, and then amplified and detected. In this connection it is to be noted that the RF probe 3, though being of a transmit/receive-combined type, may be separated into a transmit-and a receive-only probe.

DEPR:
A data acquisition unit 10 converts the MR signal which is received by the receiver 8 into a digital signal and, after being temporarily stored, transferred one at a time to a computer 11. The computer 11 has, in addition to a function of serving as a center for controlling a system as a whole, a function of creating an MR image from an inside of the subject on the basis of the MR signal, a function of calculating an absolute temperature on the basis of the MR signal, a function of calculating a relative temperature on the basis of the MR signal, a function of converting the relative temperature into the absolute temperature, a function of creating display data on the absolute temperature and relative temperature and outputting it to a display 13 and a function of detecting a motion error on the basis of a temporal variation corresponding to a resonance frequency of a fat's proton not revealing any temperature dependence.

DEPR:
A sequence controller 9 keeps pulse sequence data for imaging, pulse sequence data for measuring the absolute temperature, and pulse sequence data for measuring the relative temperature. The three kinds of these data are selectively loaded under the control of the computer and executed. The sequence controller 9 controls the gradient coil power supply 5, shim coil power supply 6, transmitter 7, receiver 8 and data acquisition unit 10. In this way, the selected pulse sequence is executed and an MR signal for measuring the absolute temperature and MR signal for measuring the relative temperature are acquired in accordance with the pulse sequence.

DEPR:
FIG. 2 shows a whole flow of a temperature monitoring operation by the present embodiment. Here it is assumed that a cerebropathia region's temperature is monitored in this case.

DEPR:
First, a subject is positioned in an imaging area at step (a). In this case it is desirable that a mechanism, such as allowing a movable type bed's top plate to slide onto a bed of the magnetic resonance apparatus, be installed so that the movable bed with the subject positioned thereon can be readily carried onto the bed on the magnetic resonance apparatus. Further, it is also desirable from the standpoint of an operation to adopt a structure with a "drip" device and other devices fixed to the top plate.

DEPR:
Then, a pulse sequence for imaging is executed by the sequence controller 9 at step (b). This pulse sequence comprises a spin echo method, or a field echo method, adopting a two-dimensional Fourier transformation (2DFT) method for example. By doing so it is possible to obtain an MR image of a cross-section including the cerebropathia region corresponding to a temperature monitoring target.

DEPR:
A therapeutic planning is determined by utilizing the MR image. At this time, it is desirable that, from the standpoint of reducing the measuring time, an MR signal be observed from one restricted temperature monitoring area. Since, however, there is a case where the brain disease of the subject is spreading or there is no brain edema or infarction focus, etc., at this stage of observation, it is necessary from the standpoint of effecting better temperature control to designate a plurality of temperature monitoring regions, obtain MR signals from the temperature monitoring regions and measure the internal temperature at each temperature monitoring region. The temperature monitoring regions are designated by an operator on the MR image, as shown in FIGS. 3A and 3B, which is obtained at step (b).

DEPR:
In the case where the hypothermia is applied, a circulation pump, cooling mat, monitoring catheter, etc., are properly prepared and set, provided that it is necessary to drive the circulation pump, etc., by an ultrasonic motor, etc., to prevent any adverse influence from the static field or to install a pump body at a location remote from a static magnet body.

DEPR:
A temperature monitoring operation--steps (c) to (i)--is started in synchronization with the start of the hypothermia. The "monitoring" is defined as a sequential observation by an operator of the temperature at the temperature monitoring regions and confirmation as to whether or not the temperature is continued at a desired level. Further, the "temperature monitoring apparatus" is defined as an apparatus for providing information necessary to such a monitoring operation, that is, presenting the temperature and time variation to the operator.

DEPR:
(Flow of the Temperature Monitoring Operation)

DEPR:
First, the pulse sequence for measuring the absolute temperature is executed at step (c).

DEPR:
Then, at step (d), a proton spectrum is calculated by the Fourier transformation from one set of MR signals collected at step (c) and the absolute temperature of the temperature monitoring regions is calculated based on the proton spectrum.

DEPR:
At step (e), the calculated absolute temperature is displayed on the display 13. Or it is possible to provide a plotter output as in the case of the outputs of measuring devices, such as an electrocardiograph, hemomanometer, electroencephalograph and venous blood saturation level meter.

DEPR:
At step (f), the pulse sequence for relative temperature measurement is executed after the absolute temperature has been measured.

DEPR:
At step (g), the relative temperature, that is, the extent to which a current temperature varies from an immediately previous temperature, is calculated based on the phase information on the extent to which the phases of the MR signals collected at step (f) vary from their immediately previous phases.

DEPR:
Then, at step (h), the relative temperature calculated at step (g) is added to the absolute temperature calculated at step (d) to convert the relative temperature into an absolute temperature. The converted absolute temperature is displayed as a numeral value or as a temperature variation graph on the display 13 or outputted from the plotter, not shown--step (i).

DEPR:
Such steps (f) to (i) are repeated until the medical treatment is finished. By doing so, the relative temperatures are sequentially measured and sequentially converted

into an absolute temperature so that the absolute temperature is constantly displayed.

DEPR:
(Pulse Sequence for Absolute Temperature measure--Frequency Method)

DEPR:
FIG. 4A shows a pulse sequence for measuring the absolute temperature. In this pulse sequence, an RF field pulse of a flip angle .alpha..degree. smaller than 90.degree. is applied together with a slice select gradient field pulse and, subsequently, a gradient field pulse Ge1 for phase encoding relating to a first axis and gradient field pulse Ge2 for phase encoding relating to a second axis are applied. In the word, a phase encode is executed with respect to each of axial directions (Ge1, Ge2) for adding special position information to the MR signals. Thereafter, an MR signal (FID signal) is generated and observed.

DEPR:
The scan time of the pulse sequence for measuring the absolute temperature is given by

DEPR:
(Method for Calculating the Absolute Temperature)

DEPR:
FIG. 4B shows a proton spectrum found by the frequency analysis of one set of MR signals collected at the pulse sequence of FIG. 4A. The chemical shift difference between the proton in H.sub.2 O and that in fat is 3.38 ppm at 27.degree. C., this corresponding to about 216 Hz under 1.5 teslas. The resonance frequency of the proton in H.sub.2 O involves a variation of -0.01 ppm per 1.degree. C. while no such temperature dependence is displayed for the proton in the fat. Therefore, with the fat set as a reference, the absolute temperature can be calculated from the chemical shift difference between the proton in the H.sub.2 O and that in the fat.

DEPR:
Since the temperature dependence of the proton in the H.sub.2 O is very small, it is necessary to obtain a proton spectrum with a high frequency resolution. To this end, it is necessary to extend the signal observation time. Further, it is considered from the standpoint of the S/N ratio that the signal observation is ended with a time of about T2* and, by the application of a zero insertion to that data, a necessary number of data is obtained to provide an apparently enhanced frequency resolution. It is also effective to enhance the measuring accuracy of the temperature through an ultra-resolution obtained by the spectrum data processing such as a non-linear least squares method.

DEPR:
Since, in the pulse sequence for measuring the absolute temperature, it is possible to, for the reduction of the measuring time, obtain only absolute temperature data on a relatively small number of temperature monitoring regions or on a relatively rough matrix, it is not always possible to measure the absolute temperature on all positions at which relative temperatures are measured by the phase method. Or it may be considered to take a procedure, such as converting those relative temperatures into absolute temperatures only at locations at which an absolute temperature is measured by the frequency method or displaying only a temperature variation at these regions. For this reason, those locations or sites (number of voxels) at which the absolute temperature is measured with the frequency method set at the step (b) for improved therapeutic results become important.

DEPR:
(Pulse Sequence for Measuring the Relative Temperature--Phase Method)

DEPR:
In the pulse sequence for measuring the relative temperature, as shown in FIG. 5, an RF field pulse of a flip angle .alpha..degree. smaller than 90.degree. is applied together with a slice select gradient field pulse Gs and, subsequently, a gradient field pulse Ge for phase encoding relating to a first axis is applied. Thereafter, an MR signal (echo) is generated by a polarity-alternated gradient field pulse Gr for frequency encoding. In the word, a phase encode is executed with respect to at least one (Ge) of axial directions for adding special position information to the MR signals, a frequency encode is executed with respect to the remaining axial direction (Gs). Those MR signals are collected during the continuation of the gradient field pulse Gr for frequency encoding.

DEPR:
Here, the scan time of the pulse sequence for measuring the relative temperature is

given by

DEPR:
Further the condition "TRa >TRb" generally is set, the scan time of the pulse sequence for measuring the relative temperature can be made shorter than that of the pulse sequence for absolute temperature measurement as shown in FIG. 4A. This reason, if being briefly explained, is as follows. The absolute temperature is measured from the frequency information contained in a magnetic resonance signal and, for the pulse sequence for measuring the absolute temperature, the frequency encoding technique cannot be used for the purpose of specifying or localizing a site. Therefore, in order to specify the site on two axes, it is necessary to effect the phase encoding with respect to the two axes. On the other hand, the relative temperature can be measured from the phase information of the MR signal. For this reason, the pulse sequence for the relative temperature can use the frequency encoding technique so as to specify the site. The phase encoding technique and frequency encoding technique can be jointly used so as to specify the site on the two axes. As well known, the pulse sequence using the phase encoding technique requires a time of the repetition time .times. the number of phase encoding steps. Therefore, the pulse sequence for the absolute temperature effects the phase encoding with respect to the two axes and this scan time becomes a time of the repetition time .times. both the number of phase encoding steps on the first axis and that of phase encoding steps on the second axis. Further, for the pulse sequence for the relative temperature, the phase encoding has only to be effected with respect to the first axis and this scan time becomes a time of the repetition time (TR) .times. the number of phase encoding steps on the first axis. For this reason, the scan time for the relative temperature can be made prominently shorter than that for absolute temperature measurement.

DEPR:
(Method for Measuring the Relative Temperature)

DEPR:
A relative temperature, that is, a temperature variation between a time point of a current pulse sequence for relative temperature measurement and a time point of an immediately previous pulse sequence of relative temperature measurement can be calculated, by the following equation, based on a variation (phase difference) between the phase of an MR signal acquired at a current pulse sequence for relative temperature measurement and that of an MR signal acquired at an immediately previous pulse sequence for relative temperature measurement.

DEPR:
FIG. 6A shows a flow of a whole pulse sequence. In FIG. 6A, "A" shows a pulse sequence for absolute temperature measurement and "B" a pulse sequence for relative temperature measurement. As shown in FIG. 6A, according to the present embodiment, a pulse sequence for relative temperature measurement is executed and, subsequently, a pulse sequence for relative temperature measurement is repeatedly executed.

DEPR:
As well-known, in the hypothermia method, the brain temperature is lowered from the start of treatment, through a continued passage of a few hours, to about 4.degree. C. The temperature control at its derivation period and re-warming period becomes very important. Since, in the method (frequency method) for calculating the absolute temperature from a H.sub.2 O proton/fat proton chemical shift difference, a scan time of its pulse sequence is relatively long as set out above, the internal temperature cannot be repeatedly measured with a practical indefinite that is short temporal resolution.

DEPR:
In the present embodiment, after the absolute temperature measuring pulse sequence (A) of a longer scan time, the relative temperature measuring pulse sequence of a short scan time is repeatedly executed. Since the relative temperature shows a temperature variation, if an absolute temperature at a given time is known, it follows that, by sequentially adding those associated relative temperatures to that absolute temperature at that given time, the absolute temperature at a subsequent time can be measured with a short temporal resolution.

DEPR:
It is to be noted that the subject, passing through the derivation period of the hypothermia, is placed under a subsequent around-the-clock intensive care and, in order to control the blood pressure, respiration, intracranial pressure and "drip", bio-measurement necessary to the above control is carried out. Although constantly effecting such temperature control is desirable during the intervening time, since the temperature measurement is required during other derivation and re-warming periods of the subject, the subject is withdrawn out of the magnetic resonance apparatus in the case where the internal temperature as well as the state of the

.subject has to be confir[redacted]. And the subject has to be ca[redacted]d into the intensive care room, etc. It is difficult to calculate an accurate t[redacted]perature variation, by the phase method, due to an adverse influence by an inhomogeneous magnetic field resulting from a positional displacement of the subject involved at the insertion and withdrawal of the subject into and out of the magnetic resonance apparatus. In the case where such an internal temperature measurement is interrupted, an absolute temperature is again measured by the frequency method and then a relative temperature is measured by the phase method to convert the relative temperature into the absolute temperature.

DEPR:
There are sometimes the cases where, during the monitoring of the internal temperature while effecting the repeated measurement of the relative temperature by the repetition of the phase method and repeated conversion of the relative temperature to the absolute temperature, a measuring temperature involves an error resulting from the motion of the subject. FIG. 6B shows a flow of a pulse sequence for realizing the detection of a motion error and initialization of the absolute temperature. When a designation is made either periodically or by an operator during the repetition of the relative temperature measuring pulse sequence ("B"), the interruption of the absolute temperature measuring pulse sequence ("A") is executed.

DEPR:
FIG. 7A shows a proton spectrum found through the application of a frequency analysis of those MR signals acquired at the absolute temperature measuring pulse sequence at time t.sub.0 in FIG. 6B. FIG. 7B shows a proton spectrum found through the application of the frequency analysis of the MR signals acquired at the absolute temperature measuring pulse sequence in time t.sub.1 in FIG. 6B. The resonance frequency of the proton in fat was .upsilon..sub.0. Let it be assumed that, at time t.sub.1, the resonance frequency of the proton in the fat varies to .upsilon..sub.1. It is considered that, since the resonance frequency of the proton in the fat reveals no temperature dependence, the time variation of the resonance frequency of the proton in the fat has a high possibility of the internal temperature monitoring region being displaced due to the motion of the subject so that a magnetic environment around it varies.

DEPR:
When the time variation (.upsilon..sub.0 -.upsilon..sub.1) of the resonance frequency has becomed too great to be disregarded and the difference (.upsilon..sub.0 -.upsilon..sub.1) has exceeded a specified threshold value, then a motion error is detected, displaying a message showing that the relative temperature measured during a time period t.sub.0 -t.sub.1 and an absolute temperature converted therefrom have a lower reliability level.

DEPR:
Further, the temperature on the basis of which the relative temperature is converted to the absolute temperature is replaced by the absolute temperature at time t.sub.1. During a treatment time as long as 24 hours, a calibration is made by utilizing the frequency method so that no motion error will grow in a cumulative way. It is, therefore, possible to continue an internal temperature monitoring with high accuracy. At such a calibration time or in the cases where a plurality of subjects are to be measured in an alternate way, it is necessary to interrupt the measurement and it is desired from the standpoint of the condition control that those graphs representing sequential temperature variations acquired both before and after such interruption be joined as a continuous one.

DEPR:
As set out above, according to the present embodiment, the temperature of the injury or the others in the subject during the hypothermia can be measured, as an absolute temperature, with a shorter temporal resolution.

DEPL:
where N denotes the number of phase encoding steps relating to the first axis in the same way as the pulse sequence for measuring the absolute temperature and TR.sub.b (.apprxeq.TR.sub.a) denotes a repetition time. This time is shorter than the scan time of the pulse sequence for absolute temperature measurement, that is, shorter than the latter scan time by an extent to which no phase encoding is done with respect to the second axis.

DEPV:
.alpha.: a coefficient showing a temperature dependence of a chemical shift of a proton in H.sub.2 O

DEPV:
.theta.(r).sub.before : a current phase image

CLPR:
1. A temperature monitoring apparatus comprising:

CLPR:
2. The temperature monitoring apparatus of claim 1, wherein:

CLPR:
3. The temperature monitoring apparatus of claim 1, wherein:

CLPR:
4. The temperature monitoring apparatus of claim 1, wherein the sequence controller executes as the absolute temperature measuring pulse sequence the following sequence:

CLPR:
5. The temperature monitoring apparatus of claim 1, wherein the sequence controller executes as the relative temperature measuring pulse sequence the following sequence:

CLPR:
6. The temperature monitoring apparatus of claim 1, further comprising:

CLPR:
7. A magnetic resonance apparatus, comprising:

CLPR:
8. The magnetic resonance apparatus of claim 1, wherein the sequence controller executes as the absolute temperature measuring pulse sequence the following sequence:

CLPR:
9. The magnetic resonance apparatus of claim 1, wherein the sequence controller executes as the relative temperature measuring pulse sequence the following sequence:

CLPR:
10. The magnetic resonance apparatus of claim 7, further comprising:

CLPR:
11. A method for monitoring a temperature, comprising:

CLPR:
12. The method of claim 11, wherein measuring of the absolute temperature comprises calculating the absolute temperature based on a chemical shift difference between a specific nucleus in a first molecule and the specific nucleus in a second molecule, and

CLPR:
13. The method of claim 11, wherein measuring the relative temperature comprises measuring a phase variation of the magnetic resonance signal detected from a specific nucleus in a first molecule, and

CLPR:
14. The method of claim 11, wherein executing the absolute temperature measuring pulse sequence comprises,

CLPR:
15. The method of claim 11, wherein executing repeatedly the relative temperature measuring pulse sequence comprises:

CLPR:
17. The method of claim 16, wherein performing the measuring the absolute temperature step comprises determining a new value of the absolute temperature.

CLPR:
18. The method of claim 16, further comprising after the measuring the absolute temperature step:

CLPR:
19. A method for monitoring a temperature comprising:

CLPR:
20. The method of claim 19, further comprising after the measuring the absolute

temperature step:

CLPV:
a gradient magnetic field generator configured to generate a gradient magnetic field;

CLPV:
a detector configured to detect a magnetic resonance signal from a subject;

CLPV:
a sequence controller configured to control the radio frequency magnetic field generator, the gradient magnetic field generator, and the detector, and configured to execute an absolute temperature measuring pulse sequence and repeatedly execute a relative temperature measuring pulse sequence following the absolute temperature measuring pulse sequence; and

CLPV:
the first processing mechanism calculates the absolute temperature based on a chemical shift difference between a specific nucleus in a first molecule and the specific nucleus in a second molecule, and

CLPV:
a first magnetic resonance frequency of the first molecule is temperature dependent, and a second magnetic resonance frequency of the second molecule is not temperature dependent.

CLPV:
the first processing mechanism calculates the relative temperature based on a phase variation of the magnetic resonance signal detected from a specific nucleus in a first molecule, and

CLPV:
a magnetic resonance frequency of the first molecule is temperature dependent.

CLPV:
a radio frequency magnetic field pulse applied concurrently with a slice select gradient magnetic field pulse, followed by

CLPV:
a first encoding gradient magnetic field pulse applied concurrently with a second encoding gradient magnetic field pulse,

CLPV:
the first encoding gradient magnetic field pulse providing phase encoding and being executed with respect to a first axial direction,

CLPV:
the second encoding gradient magnetic field pulse providing phase encoding and being executed with respect to a second axial direction, and

CLPV:
the first encoding gradient magnetic field pulse and the second encoding gradient magnetic field pulse adding position information to the magnetic resonance signal detected from the subject.

CLPV:
a radio frequency magnetic field pulse applied concurrently with a slice select gradient magnetic field pulse, followed by

CLPV:
a phase encoding gradient magnetic field pulse applied concurrently with a frequency encoding gradient magnetic field pulse,

CLPV:
the phase encoding gradient magnetic field pulse providing phase encoding and being executed with respect to a first axial direction and adding position information to the magnetic resonance signal detected from the subject, and

CLPV:
the frequency encoding gradient magnetic field pulse being executed with respect to a second axial direction.

CLPV:
wherein a magnetic resonance frequency of the specific molecule is not temperature

t.

radient magnetic field generator configured to generate a gradient magnetic field;

CLPV:
a detector configured to detect a magnetic resonance signal from a subject;

CLPV:
a sequence controller configured to control the radio frequency magnetic field generator, the gradient magnetic field generator, and the detector, and configured to execute an absolute temperature measuring pulse sequence and subsequently, to execute repeatedly a relative temperature measuring pulse sequence;

CLPV:
a first processing mechanism configured to calculate the absolute temperature of a region of interest in the subject based on frequency information of the magnetic resonance signal detected from the subject in response to the absolute temperature measuring pulse sequence; and

CLPV:
a second processing mechanism configured to calculate the relative temperature of the region of interest based on phase information of a magnetic resonance signal detected from the subject in response to the relative temperature measuring pulse sequence.

CLPV:
a radio frequency magnetic field pulse applied concurrently with a slice select gradient magnetic field pulse, followed by

CLPV:
a first encoding gradient magnetic field pulse applied concurrently with a second encoding gradient magnetic field pulse,

CLPV:
the first encoding gradient magnetic field pulse providing phase encoding and being executed with respect to a first axial direction,

CLPV:
the second encoding gradient magnetic field pulse providing phase encoding and being executed with respect to a second axial direction, and

CLPV:
the first encoding gradient magnetic field pulse and the second encoding gradient magnetic field pulse adding position information to the magnetic resonance signal detected from the subject.

CLPV:
a radio frequency magnetic field pulse applied concurrently with a slice select gradient magnetic field pulse, followed by

CLPV:
a phase encoding gradient magnetic field pulse applied concurrently with a frequency encoding gradient magnetic field pulse,

CLPV:
the phase encoding magnetic field pulse providing phase encoding and being executed with respect to a first axial direction, and adding position information to the magnetic resonance signal detected from the subject, and

CLPV:
the frequency encoding gradient magnetic field pulse being executed with respect to a second axial direction.

CLPV:
executing an absolute temperature measuring pulse sequence;

CLPV:
measuring the absolute temperature of a region of interest in a subject based on frequency information of a magnetic resonance signal detected from the subject in response to the absolute temperature measuring pulse sequence; and

CLPV:
determining repeatedly a corresponding absolute temperature from a relative

temperature, including

CLPV:
a first magnetic resonance frequency of the first molecule is temperature dependent, and a second magnetic resonance of the second molecule is not temperature dependent.

CLPV:
a magnetic resonance frequency of the first molecule is temperature dependent.

CLPV:
applying a radio frequency magnetic field pulse concurrently with a slice select gradient magnetic field pulse, followed by

CLPV:
applying a first encoding gradient magnetic field pulse concurrently with a second encoding gradient magnetic field pulse,

CLPV:
the first encoding gradient magnetic field pulse providing phase encoding and being executed with respect to a first axial direction,

CLPV:
the second encoding gradient magnetic field pulse providing phase encoding and being executed with respect to a second axial direction, and

CLPV:
the first encoding gradient magnetic field pulse and the second encoding gradient magnetic field pulse adding position information to the magnetic resonance signal detected from the subject.

CLPV:
applying a radio frequency magnetic field pulse concurrently with a slice select gradient magnetic field pulse, followed by

CLPV:
applying a phase encoding gradient magnetic field pulse concurrently with a frequency encoding gradient magnetic field pulse,

CLPV:
the phase encoding magnetic field pulse providing phase encoding and being executed with respect to a first axial direction, and adding position information to the magnetic resonance signal detected from the subject, and

CLPV:
the frequency encoding gradient magnetic field pulse being executed with respect to a second axial direction.

CLPV:
performing the executing the absolute temperature measuring pulse sequence step;

CLPV:
performing the measuring the absolute temperature step; and

CLPV:
calculating a magnetic resonance frequency based on the magnetic resonance signal detected from a specific nucleus in a specific molecule; and

CLPV:
detecting a motion error of the subject based on a temporal variation of the calculated magnetic resonance frequency,

CLPV:
a magnetic resonance frequency of the specific molecule not being temperature dependent.

CLPV:
executing an absolute temperature measuring pulse sequence;

CLPV:
measuring the absolute temperature of a region of interest in a subject based on frequency information of a magnetic resonance signal detected from the subject in response to the absolute temperature measuring pulse sequence;

CLPV:

determining repeatedly a corresponding absolute temperature from a relative temperature, including

CLPV:
calculating a magnetic resonance frequency based on the magnetic resonance signal detected from a specific nucleus in a specific molecule; and

CLPV:
detecting a motion error of the subject based on a temporal variation of the calculated magnetic resonance frequency,

CLPV:
a magnetic resonance frequency of the specific molecule not being temperature dependent.

CLPW:
to calculate the absolute temperature of a region of interest in the subject based on frequency information of the magnetic resonance signal detected from the subject in response to the absolute temperature measuring pulse sequence,

CLPW:
to calculate the relative temperature of the region of interest in the subject based on phase information of the magnetic resonance signal detected from the subject in response to the relative temperature measuring pulse sequence, and

CLPW:
to convert the measured relative temperature to a corresponding absolute temperature based on the calculated absolute temperature.

CLPW:
to calculate a magnetic resonance frequency based on the magnetic resonance signal detected from a specific nucleus in a specific molecule, and

CLPW:
to detect a motion error of the subject based on a temporal variation of the calculated magnetic resonance frequency;

CLPW:
to calculate a magnetic resonance frequency based on the magnetic resonance signal detected from a specific nucleus in a specific molecule, and

CLPW:
to detect a motion error of the subject based on a temporal variation of the calculated magnetic resonance frequency,

CLPW:
a magnetic resonance frequency of the specific molecule is not temperature dependent.

CLPW:
executing a relative temperature measuring pulse sequence, including,

CLPW:
measuring the relative temperature of the region of interest in the subject based on phase information of the magnetic resonance signal detected from the subject in response to the relative temperature measuring pulse sequence, and

CLPW:
converting the relative temperature into the corresponding absolute temperature based on the absolute temperature.

CLPW:
executing a relative temperature measuring pulse sequence

CLPW:
measuring the relative temperature of the region of interest in the subject based on phase information of the magnetic resonance signal detected from the subject in response to the relative temperature measuring pulse sequence, and

CLPW:
converting the relative temperature into the corresponding absolute temperature based on the absolute temperature; and

CLPW:

the executing the absolute temperature measuring pulse sequence step,

CLPW:
the measuring the absolute temperature step, and

ORPL:
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ORPL:
Yasutoshi Ishihara, et al. "A Precise and Fast Temperature Mapping Using Water Proton Chemical Shift", Magnetic Resonance In Medicine, vol. 34, 1995, pp. 814-823

ORPL:
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☐ 1. Document ID: US 6194899 B1

L10: Entry 1 of 3 File: USPT Feb 27, 2001

US-PAT-NO: 6194899
DOCUMENT-IDENTIFIER: US 6194899 B1

TITLE: Temperature monitoring method, temperature monitoring apparatus and magnetic resonance apparatus

DATE-ISSUED: February 27, 2001

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US-CL-CURRENT: 324/315; 600/412

Full | Title | Citation | Front | Review | Classification | Date | Reference | KWIC | Draw Desc | Image

☐ 2. Document ID: US 6054855 A

L10: Entry 2 of 3 File: USPT Apr 25, 2000

US-PAT-NO: 6054855
DOCUMENT-IDENTIFIER: US 6054855 A

TITLE: Magnetic susceptibility control of superconducting materials in nuclear magnetic resonance (NMR) probes

DATE-ISSUED: April 25, 2000

INVENTOR-INFORMATION:

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US-CL-CURRENT: 324/318; 324/307

Full | Title | Citation | Front | Review | Classification | Date | Reference | KWIC | Draw Desc | Image

☐ 3. Document ID: US 4689563 A

L10: Entry 3 of 3 File: USPT Aug 25, 1987

US-PAT-NO: 4689563
DOCUMENT-IDENTIFIER: US [redacted]9563 A

TITLE: High-field nuclear magnetic resonance imaging/spectroscopy system

DATE-ISSUED: August 25, 1987

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US-CL-CURRENT: 324/309

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L9: Entry 3 of 4 File: USPT Jan 3, 1995

DOCUMENT-IDENTIFIER: US 5378987 A
TITLE: Method and apparatus for non-invasive measurement of temperature distribution within target body using nuclear magnetic resonance imaging

ABPL:
A non-invasive measurement of a temperature distribution within a target body using a nuclear magnetic resonance imaging, capable of realizing a high speed and a high precision measurement, and accounting for a displacement of the target body during the measurement. The chemical shift data from the target body at each voxel in an imaging target region on the target body are collected with and without a temperature change of the target body, a difference between the chemical shift data collected with the temperature change and the chemical shift data collected without the temperature change at each voxel, and a temperature distribution image is constructed and displayed according to the difference calculated. The chemical shift data are preferably collected by using a phase mapping imaging sequence.

BSPR:
The present invention relates to a non-invasive measurement of a temperature distribution within a target body, which utilizes a nuclear magnetic resonance (NMR) imaging technique.

BSPR:
In recent years, a need for developing a method for measuring a temperature distribution within a target body such as a living body non-invasively has been felt strongly in a wide range of medical fields including temperature measurement, tissue temperature measurement, and hyperthermia treatment.

BSPR:
This need is motivated by the fact that the temperature within the living body is a physical quantity which reflects many physiological functions of the living body, so that the information on the temperature distribution can be useful in diagnosing diseases such as circulation malfunction and tumors, as well as in monitoring the temperature change during the heating process used in hyperthermia treatment.

BSPR:
To this end, there has been several propositions to utilize the temperature dependent parameters of the NMR signals, such as a thermal equilibrium magnetization, longitudinal relaxation time, transverse relaxation time, and diffusion constant, for the purpose of the non-invasive measurement of the temperature distribution within a target body, which include the following representative cases.

BSPR:
The thermal equilibrium magnetization M.sub.0 is known to be inversely proportional to the temperature, as expressed by the following expression (1).

BSPR:
According to this expression (1), the temperature gradient of the thermal equilibrium magnetization M.sub.0 for the proton system in pure water is -0.36%/K at the temperature of 40.degree. C., so that the temperature can be estimated from the change of the thermal equilibrium magnetization M.sub.0.

BSPR:
However, the temperature gradient of the thermal equilibrium magnetization M.sub.0 takes a very small value, and the measurement must be made on a basis of the proton density image obtained by the NMR imaging, so that the very high precision measurement is required in order to achieve the sufficient temperature resolution and accuracy.

BSPR:
Here, the temperature gradient is 2.2%/K for the proton system in pure water at the temperature of 40.degree. C., which is larger than the temperature gradient of the

thermal equilibrium magnetization M.sub.0. Thus, the longitudinal relaxation time T.sub.1 is a thermally more sensitive parameter than the thermal equilibrium magnetization M.sub.0.

BSPR:
However, the use of this longitudinal relaxation time T.sub.1 for the temperature measurement has been associated with the following problems.

BSPR:
First, it is necessary to measure the temperature dependency of each tissue in advance, because the ratio of the free water and the bound water and the difference of the viscosity affect the temperature dependency.

BSPR:
The transverse relaxation time can be expressed as a function of temperature, as in the following expressions (4) and (5).

BSPR:
However, the use of this transverse relaxation time T.sub.2 for the temperature measurement has been associated with the following problems.

BSPR:
First, it is necessary to measure the temperature dependency of each tissue in advance, because the ratio of the free water and the bound water and the difference of the viscosity affect the temperature dependency.

BSPR:
The diffusion constant is known to have the temperature dependency expressed by the following expression (6).

BSPR:
According to this expression (6), the temperature change (T-T.sub.0) can be determined from the diffusion constants D.sub.0 and D obtained before and after the temperature change, by the following expression (7).

BSPR:
However, such a use of this diffusion constant D for the temperature measurement is based on assumptions that (T-T.sub.0)<
BSPR:
Furthermore, in such a conventional non-invasive measurement of the temperature distribution within a target body utilizing the temperature dependent parameters of the NMR signals, there has been a possibility for the measured temperature distribution to be spoiled by the error due to a displacement of the target body during the temperature measurement, especially when the target body is a living body, because the temperature dependency of the chemical shifts is substantially smaller compared with an inhomogeneity of the magnetic field caused by a body movement of the target body.

BSPR:
It is therefore an object of the present invention to provide a method and an apparatus for non-invasive measurement of a temperature distribution within a target body using a nuclear magnetic resonance imaging, capable of measuring the temperature distribution within a target body at a high speed and a high precision, and visualizing the measured temperature distribution.

BSPR:
It is another object of the present invention to provide a method and an apparatus for non-invasive measurement of a temperature distribution within a target body using a nuclear magnetic resonance imaging, capable of measuring the temperature distribution within a target body while accounting for a displacement of the target body during the temperature measurement.

BSPR:
According to one aspect of the present invention there is provided an apparatus for measuring temperature distribution in a target body, comprising: nuclear magnetic resonance imaging means for collecting chemical shift data from the target body at each voxel in an imaging target region on the target body, with and without a temperature change of the target body; calculation means for calculating a difference between the chemical shift data collected with the temperature change and the chemical shift data collected without the temperature change at each voxel; image construction means for constructing a temperature distribution image according to the difference calculated by the calculation means at each voxel: and display means for displaying the temperature distribution image constructed by the image construction means.

BSPR:
According to another aspect of the present invention there is provided a method of measuring temperature distribution in a target body, comprising the steps of: (a) collecting chemical shift data from the target body at each voxel in an imaging target region on the target body by nuclear magnetic resonance imaging means, with and without a temperature change of the target body; (b) calculating a difference between the chemical shift data collected at the step (a) with the temperature change and the chemical shift data collected at the step (a) without the temperature change at each voxel; (c) constructing a temperature distribution image according to the difference calculated at each voxel at the step (b); and (d) displaying the temperature distribution image constructed at the step (c).

BSPL:
where T is an absolute temperature, N.sub.0 is a proton density, .gamma. is a gyromagnetic ratio, h is a Planck constant, k is a Boltzmann constant, and B.sub.0 is a static magnetic field strength.

DRPR:
FIG. 1 is a schematic block diagram of one embodiment of a nuclear magnetic resonance imaging apparatus having a non-invasive temperature distribution measurement function according to the present invention.

DRPR:
FIG. 2 is an imaging sequence diagram for a 4D-MRSI (4-Dimensional Magnetic Resonance Spectroscopic Imaging) sequence suitable for the chemical shift measurement to be made by the apparatus of FIG. 1.

DRPR:
FIG. 3 is an illustration of an exemplary spectral data containing a temperature independent chemical shift component and a temperature dependent chemical shift component.

DRPR:
FIG. 7 is a diagrammatic illustration of an example of the imaging target view field to be used in the imaging sequence of FIG. 6 containing the temperature independent region.

DRPR:
FIG. 8 is a diagrammatic illustration of another example of the imaging target view field to be used in the imaging sequence of FIG. 6 containing the temperature independent points.

DRPR:
FIG. 9 is a diagrammatic illustration of another example of the imaging target view field to be used in the imaging sequence of FIG. 6 containing a phantom made of the temperature independent chemical shift components.

DRPR:
FIG. 10 is an illustration of an NMR image obtained by using the imaging target view field of FIG. 9 at a measurement plane indicated in FIG. 9.

DEPR:
Referring now to FIG. 1, one embodiment of a method and an apparatus for non-invasive measurement of a temperature distribution within a target body using a nuclear magnetic resonance imaging according to the present invention will be described in detail.

DEPR:
FIG. 1 schematically shows a configuration of a nuclear magnetic resonance imaging apparatus having a non-invasive temperature distribution measurement function.

DEPR:
In this configuration of FIG. 1, the apparatus comprises: a main magnet 10 for generating a static magnetic field in which a target body such as a living body as a measurement target is placed; a main magnet power source 11 for driving the main magnet 10 appropriately; gradient coil system 12 for generating gradient magnetic fields in three mutually orthogonal directions (X, Y, and Z directions) in superposition to the static magnetic field; a gradient coil power source 13 for driving the gradient coil system 12 appropriately; a shim coil system 14 for adjusting the homogeneity of the static magnetic field generated by the main magnet 10; a shim coil power source 15 for driving the shim coil system 14 appropriately; an RF (radio frequency) probe 16 for applying RF pulses to the target body and detecting NMR signals emitted from the target body in response to the application of the RF

pulses; a transmitter 17 for supplying appropriate RF pulses to the RF probe 16; a receiver 18 for receiving, amplifying, and collecting the NMR signals detected by the RF probe 16; a sequence controller 19 for controlling the gradient coil power source 13, the shim coil power source 15, and the transmitter 17 to realize appropriate imaging sequence; a CPU 20 for controlling the operation of the apparatus as a whole, as well as reconstructing the NMR images from the NMR signals collected by the receiver 18; a heating device 21 for heating the target body in the hyperthermia treatment; a heating control device 22 for driving the heating device 21 appropriately; and a display 9 for displaying the NMR images reconstructed by the CPU 20.

DEPR:
Now, the principle behind the non-invasive measurement of a temperature distribution within a target body according to the present invention will be described.

DEPR:
The present invention utilizes the known fact that the hydrogen bond strength dependent on temperature can affect the shielding constant related to the chemical shift, so that the temperature can be determined from the change of the chemical shift for the hydrogen bonded hydroxy radical (--OH). For example, it is known that the chemical shift for the hydroxy radical (--OH), such as the hydroxy radical (--OH) in pure water, the hydroxy radical (--OH) with respect to the methyl radical (--CH.sub.3) in the methanol (CH.sub.3 OH), and the hydroxy radical (--OH) with respect to the methylene radical (--CH.sub.2 --) in ethylene glycol (HO--CH.sub.2 --CH.sub.2 --OH), is proportional to the temperature by an amount of approximately -0.01 ppm/.degree.C., regardless of the chemical shift measurement target object containing the hydroxy radical (--OH). (See, J. C. Hindman, Journal of Chemical Physics, Vol. 44, No. 12, pp. 4582-4592, 1966.)

DEPR:
Thus, the temperature can be calculated from the result of the chemical shift measurement by establishing the relationship between the chemical shift data and the temperatures in advance.

DEPR:
In this embodiment, the chemical shift measurement is achieved in the apparatus of FIG. 1 by carrying out an imaging sequence called 4D-MRSI (4-Dimensional Magnetic Resonance Spectroscopic Imaging) sequence shown in FIG. 2, which is capable of obtaining the chemical shift data as well as the three-dimensional spatial data. In FIG. 2, the .alpha..degree. RF pulse 23 is applied by the RF probe 16 while the gradient magnetic fields 24, 25, and 26 in the Z, Y, and X directions, respectively, are generated by the gradient coil system 12, at the timings indicated in FIG. 2. It is also possible to utilize the imaging sequence for collecting the spectral data by successively switching the gradient magnetic fields, in order to realize a high speed data collection.

DEPR:
By using this imaging sequence of FIG. 2, the temperature change in the living body due to the heating applied by the heating device 21 can be determined from the measurements of the shifting of the spectral data for H.sub.2 O at each voxel, before and after the heating by the heating device 21.

DEPR:
Thus, when the hyperthermia treatment is carried out is by the heating device 21, the effect of the treatment can be judged non-invasively, by measuring the chemical shifts for H.sub.2 O at each voxel before and after the heating, calculating the difference between the chemical shifts at each voxel before and after the heating, and displaying the calculated differences for all voxels in a form of temperature distribution image on the display 9 in the apparatus of FIG. 1.

DEPR:
Here, in a case the measurement target portion in the target body is tissues containing .sup.1 H compounds such as brain tissues, the temperature independent chemical shift components such as CH.sub.2 and CH.sub.3 can be utilized as a reference for determining the absolute temperature, so that the temperature distribution image given in the absolute temperature can be displayed on the display 9 in the apparatus of FIG. 1. For example, in the spectral data shown in FIG. 3 which is obtained from a voxel in which the water component and the fat component are co-existing, the peak 28 for the methylene radical (--CH.sub.2 --) due to the fat component can be utilized as the reference for determining the absolute temperature for the peak 27 for the hydroxy radical (--OH) due to the water component.

DEPR:
As a result, it also becomes possible in this embodiment to diagnose the disease of

the target body from the observation of the abnormality in the temperature distribution image given in the absolute temperature.

DEPR:
Alternatively, FIG. 4 shows another imaging sequence suitable for achieving the chemical shift measurement in the apparatus of FIG. 1, which utilizes the phase mapping technique. In this imaging sequence of FIG. 4, the phase mapping technique is applied to the spin echo sequence, as an illustrative example. In FIG. 4, the 90.degree. RF pulse 29 and the 180.degree. RF pulse 30 are applied by the RF probe 16 and the NMR echo signal 31 is detected by the RF probe 16, while the gradient magnetic fields 32, 33, 34 and 35 in the Z, Y, and X directions are generated by the gradient coil system 12, at the timings indicated in FIG. 4, where the 90.degree. RF pulse 29 and the 180.degree. RF pulse 30 are separated by a period .tau., while the 180.degree. RF pulse 30 and the NMR echo signal 31 are separated by a period .tau.+.DELTA..tau..

DEPR:
Here, the phase difference .theta.(x, y, z) at each position (x, y, z) between two NMR images obtained before and after the heating by the heating device 21 with different values for the period .tau. between the 90.degree. RF pulse 29 and the 180.degree. RF pulse 30 can be expressed by the following expression (8).

DEPR:
On the other hand, it is also possible to obtain the temperature change .DELTA.T in which the influences of the inhomogeneity of the static magnetic field (.delta.B.sub.H) and the chemical shift difference (.delta.B.sub.C) are removed, from the measurements of the phase difference distribution .theta..sub.0 (x, y, z) before the heating and the phase difference distribution .theta..sub.1 (x, y, z) after the heating, according to the following expression (9).

DEPR:
Thus, by visualizing the temperature change .DELTA.T in a form of a phase difference image according to this equation (9) and displaying the phase difference image on the display 9 in the apparatus of FIG. 1, it also becomes possible to judge the effect of the hyperthermia treatment carried out by the heating device 21, non-invasively, just as in a case of using the imaging sequence of FIG. 2 described above.

DEPR:
Here, it is to be noted that, in a case the measurement target portion in the target body is tissues containing .sup.1 H compounds such as brain tissues, the phase mapping image of the temperature independent chemical components such as CH.sub.2 and CH.sub.3 alone can be obtained by using the appropriate RF pulse sequences for selectively exciting or saturating only these temperature independent chemical shift components prior to the imaging sequence of FIG. 4, such that the temperature distribution image given in the absolute temperature can be calculated from difference between the phase mapping image for H.sub.2 O and the phase mapping image for these temperature independent chemical shift components, and displayed on the display 9 in the apparatus of FIG. 1.

DEPR:
As a result, it also becomes possible to diagnose the disease of the target body from the observation of the abnormality in the temperature distribution image given in the absolute temperature, just as in a case of using the imaging sequence of FIG. 2 described above.

DEPR:
It is also to be noted that, in a case of using this imaging sequence of FIG. 4, any desired dynamic range can be set up by appropriately adjusting the value of .DELTA..tau. indicated in FIG. 4 according to the level of the temperature change to be measured, so that the precision in the measurement can be improved.

DEPR:
It is also possible for this embodiment to utilize the imaging sequence shown in FIG. 5 in which the phase mapping technique is applied to the field echo sequence, as an illustrative example. In FIG. 5, the .alpha..degree. RF pulse 36 is applied by the RF probe 16 while the gradient magnetic fields 40 and 41 in the Z and Y directions are generated by the gradient coil system 12, and then the NMR echo signals 37, 38, and 39 are successively detected by the RF probe 16 while the gradient magnetic field 42 in the X direction which is successively switching its polarity is generated by the gradient coil system 12, such that a high speed data collection can be realized by the successive acquisition of the NMR echo signals 37, 38, and 39 in response to the successive switching of the polarity of the gradient magnetic field 42.

DEPR:

In this case, the expressions (8) and (9) described above could be rewritten in terms of a period TE between the .alpha..degree. RF pulse and the NMR echo signal 37 as the following expressions (10) and (11), respectively.

DEPR:
It is to be noted that, in this embodiment, any desired number of the temperature independent chemical shift components may be used as references for determining the absolute temperature. In particular, the use of a plurality of the references is preferable in achieving the higher accuracy in the absolute temperature determination. In addition, the temperature independent chemical shift components located outside of the target body may also be utilized for this purpose.

DEPR:
It is also to be noted that the temperature distribution image of the target body obtained by this embodiment can be utilized for the diagnosis of the diseases such as the circulation malfunction and tumors.

DEPR:
Moreover, the temperature distribution image of the target body obtained by this embodiment may also be utilized in controlling the heating regions and the heating levels in the hyperthermia treatment.

DEPR:
Furthermore, it is also possible to correct the deviation of the chemical shift due to the temperature change by utilizing a plurality of pH dependent chemical shift radicals, from which the pH distribution in the target body can be obtained by using the NMR imaging.

DEPR:
As described, according to this embodiment, it becomes possible to provide a method and an apparatus for non-invasive measurement of a temperature distribution within a target body using a nuclear magnetic resonance imaging, capable of measuring the temperature distribution within a target body at a high speed and a high precision, and visualizing the measured temperature distribution.

DEPR:
Now, as already mentioned above, the temperature dependency of the chemical shifts is as small as an order of 0.01 ppm, which is substantially smaller compared with an inhomogeneity of the magnetic field that can be caused by a body movement of the target body, so that there is a possibility for the measured temperature distribution to be spoiled by the error due to a displacement of the target body during the temperature measurement, especially when the target body is a living body. This problem of the error due to a displacement of the target body can be resolved according to the present invention as follows.

DEPR:
Namely, the phase mapping images of a temperature independent region containing the temperature independent chemical shift components are also obtained before and after the heating, so as to detect the phase change due to the body movement of the target body according to these phase mapping images.

DEPR:
In FIG. 6, in the displacement detection part, the RF pulse 50 (60) is applied by the RF probe 16 while the gradient magnetic fields 51 (61), 52 (62), and 53 (63) in the Z, Y, and X directions are generated by the gradient coil system 12 at timings indicated in FIG. 6, and then the NMR echo signals 54 (64) are successively detected by the RF probe 16 while the gradient magnetic field 53 (63) in the X direction which is successively switching its polarity is generated by the gradient coil system 12, such that an ultra high speed data collection can be realized by the successive acquisition of the NMR echo signals 54 (64) in response to the successive switching of the polarity of the gradient magnetic field 53 (63).

DEPR:
Then, in the chemical shift measurement part, the RF pulse 55 (65) is applied by the RF probe 16 while the gradient magnetic fields 56 (66), 57 (67), and 58 (68) in the Z, Y, and X directions are generated by the gradient coil system 12 at timings indicated in FIG. 6, and then the NMR echo signals 59 (69) are successively detected by the RF probe 16 while the gradient magnetic field 58 (68) in the X direction which is successively switching its polarity is generated by the gradient coil system 12, such that an ultra high speed data collection can be realized by the successive acquisition of the NMR echo signals 59 (69) in response to the successive switching of the polarity of the gradient magnetic field 58 (68).

DEPR:

The second sequence differs from the first sequence in that the echo time to be used in the chemical shift measurement part is displaced from the first sequence to the second sequence by .DELTA.TE as indicated in FIG. 6, so as to obtain two sets of phase mapping images including a first phase mapping image obtained by the first sequence and a second phase mapping image obtained by the second sequence.

DEPR:
The entire imaging sequence shown in FIG. 6 is executed both before and after the heating. The phase difference at each voxel between the first phase mapping image and second phase mapping image is calculated in each execution of the imaging sequence of FIG. 6. Then, the difference image is calculated from the calculated phase differences obtained from the execution before the heating and the execution after the heating, and the calculated difference image is displayed on the display 9 in the apparatus of FIG. 1. Here, two sets of phase mapping images are used in order to remove the influence of the image shift in the encoding direction in the ultra high speed imaging sequence due to the change of the magnetic field distributions before and after the heating.

DEPR:
Here, as shown in FIG. 7, the data can be collected in the displacement detection part from an entire imaging target view field VF which includes the temperature independent region TIR containing the temperature independent chemical shift components, along with the temperature dependent region TDR containing the temperature dependent chemical shift components whose chemical shifts are to be measured in the chemical shift measurement part.

DEPR:
Alternatively, as shown in FIG. 8, the data can be collected in the displacement detection part from predetermined number of temperature independent points TIP within the entire imaging target view field VF at which the temperature independent chemical shift components are located, by using a localized excitation technique. The use of the temperature independent points TIP will be advantageous in terms of a time required for the displacement detection part.

DEPR:
It is also possible, as shown in FIG. 9, to utilize a phantom 70 made of the temperature independent chemical shift components such as fat, which is to be attached to the measurement target body and placed within the entire imaging target view field VF along with the temperature dependent region TDR containing the temperature dependent chemical shift components whose chemical shifts are to be measured in the chemical shift measurement part. In this case, the NMR image taken at a measurement plane MP indicated in FIG. 9 appears as shown in FIG. 10 for example. Here, it is preferable for the phantom 70 to have a shape which can cause a large inhomogeneity to the static magnetic field when the target body moves, such that the phase change due to the body movement of the target body can be detected easily.

DEPR:
On the other hand, when the phase change due to the body movement of the target body is detected from the difference image, it is also possible to adjust the strength of the gradient magnetic fields determining the imaging target region in the target body to displace the imaging target region such that the phase change in the difference image can be nullified. In other words, by sequentially displacing the imaging target region by the appropriate adjustment of the gradient magnetic fields, it is possible to find the optimum imaging target region in which the phase change in the difference image can be nullified.

DEPR:
As described, according to this embodiment, it also becomes possible to realize a non-invasive measurement of a temperature distribution within a target body using a nuclear magnetic resonance imaging, accounting for a displacement of the target body during the temperature measurement.

DEPL:
where .gamma. is the gyromagnetic ratio and .delta.B(x, y ,z) is a spatial deviation accounting for the inhomogeneity of the static magnetic field (.delta.B.sub.H), the chemical shift difference (.delta.B.sub.C), and the temperature dependency (.delta.B.sub.T), at each position (x, y, z).

DEPL:
where .beta. is a temperature dependency coefficient.

CLPR:
1. An apparatus for measuring temperature distribution in a target body, comprising:

CLPR:
2. The apparatus of claim 1, wherein the nuclear magnetic resonance imaging means collects the chemical shift data for at least one temperature dependent chemical component in the target body after and before the temperature change of the target body, as the chemical shift data with and without the temperature change, respectively.

CLPR:
3. The apparatus of claim 1, wherein the nuclear magnetic resonance imaging means collects the chemical shift data for at least one temperature dependent chemical component and at least one temperature independent chemical shift component in the target body, as the chemical shift data with and without the temperature change, respectively.

CLPR:
4. The apparatus of claim 1, wherein the phase mapping imaging sequence used by the nuclear magnetic resonance imaging means is any one of a phase mapped spin echo sequence, a phase mapped field echo sequence, and a phase mapped ultra high speed imaging sequence in which a plurality of nuclear magnetic resonance signals are collected from the target body in response to repeated reversals of a polarity of a reading gradient magnetic field.

CLPR:
5. The apparatus of claim 1, wherein the imaging target region contains a temperature independent portion and the apparatus further comprises:

CLPR:
6. The apparatus of claim 5, wherein the nuclear magnetic resonance imaging means carries out the phase mapping imaging sequence for collecting the chemical shift data for the temperature independent portion separately from the phase mapping imaging sequence for collecting the chemical shift data for a temperature dependent portion in the imaging target region.

CLPR:
7. The apparatus of claim 5, wherein the temperature independent portion in the imaging target region is provided by a phantom made of temperature independent chemical shift components which is attached to the target body and placed inside the imaging target region.

CLPR:
9. The apparatus of claim 1, wherein the nuclear magnetic resonance imaging means collects two sets of each of the chemical shift data with and without the temperature change, by using a phase mapped echo sequence with different echo times; and wherein the calculation means takes a first phase difference between the two sets of the chemical shift data collected with the temperature change and a second phase difference between the two sets of the chemical shift data collected without the temperature change, and calculates a difference between the first phase difference and the second phase difference as the difference between the chemical shift data collected with and without the temperature change at each voxel.

CLPR:
10. A method of measuring temperature distribution in a target body, comprising the steps of:

CLPR:
11. The method of claim 10, wherein at the step (a), the nuclear magnetic resonance imaging unit collects the chemical shift data for at least one temperature dependent chemical component in the target body after and before the temperature change of the target body, as the chemical shift data with and without the temperature change, respectively.

CLPR:
12. The method of claim 10, wherein at the step (a), the nuclear magnetic resonance imaging unit collects the chemical shift data for at least one temperature dependent chemical component and at least one temperature independent chemical shift component in the target body, as the chemical shift data with and without the temperature change, respectively.

CLPR:
13. The method of claim 10, wherein at the step (a), the phase mapping imaging sequence used by the nuclear magnetic resonance imaging unit is any one of a phase mapped spin echo sequence, a phase mapped field echo sequence, and a phase mapped ultra high speed imaging sequence in which a plurality of nuclear magnetic resonance signals are collected from the target body in response to repeated reversals of a

polarity of a reading gradient magnetic field.

CLPR:
14. The method of claim 10, wherein the imaging target region contains a temperature independent portion and the method further comprises the step of:

CLPR:
15. The method of claim 14, wherein at the step (a), the nuclear magnetic resonance imaging unit carries out the phase mapping imaging sequence for collecting the chemical shift data for the temperature independent portion separately from the phase mapping imaging sequence for collecting the chemical shift data for a temperature dependent portion in the imaging target region.

CLPR:
16. The method of claim 14, wherein the temperature independent portion in the imaging target region is provided by a phantom made of temperature independent chemical shift components which is attached to the target body and placed inside the imaging target region.

CLPR:
18. The method of claim 10, wherein at the step (a), the nuclear magnetic resonance imaging unit collects two sets of each of the chemical shift data with and without the temperature change, by using a phase mapped echo sequence with different echo times; and at the step (b), a first phase difference between the two sets of the chemical shift data collected with the temperature change and a second phase difference between the two sets of the chemical shift data collected without the temperature change are taken, and a difference between the first phase difference and the second phase difference is calculated as said difference between the chemical shift data collected with and without the temperature change at each voxel.

CLPV:
nuclear magnetic resonance imaging means for collecting chemical shift data from the target body at each voxel in an imaging target region in the target body, with and without a temperature change of the target body using a phase mapping imaging sequence;

CLPV:
calculation means for calculating a difference between the chemical shift data collected with the temperature change and the chemical shift data collected without the temperature change at each voxel as a phase difference at each voxel between the chemical shift data collected with and without the temperature change;

CLPV:
image construction means for constructing a temperature distribution image according to the difference calculated by the calculation means at each voxel by converting the phase difference at each voxel calculated by the calculation means into a corresponding temperature; and

CLPV:
display means for displaying the temperature distribution image constructed by the image construction means.

CLPV:
displacement detection means for detecting a displacement of the target body between a collection of the chemical shift data with the temperature change and a collection of the chemical shift data without the temperature change according to a phase change in the chemical shift data collected from the temperature independent portion of the imaging target region with and without the temperature change.

CLPV:
means for adjusting strengths of gradient magnetic fields used by the nuclear magnetic resonance imaging means, when the displacement detection means detects the displacement, to shift the imaging target region sequentially to a position at which the phase change is nullified.

CLPV:
(a) collecting chemical shift data from the target body at each voxel in an imaging target region in the target body with a nuclear magnetic resonance imaging unit, with and without a temperature change of the target body, using a phase mapping imaging sequence;

CLPV:
(b) calculating a difference, between the chemical shift data collected at the step (a) with the temperature change and the chemical shift data collected at the step (a)

without the temperature change at each voxel, as a phase difference at each voxel between the chemical shift data collected with and without the temperature change;

CLPV:
(c) constructing a temperature distribution image according to the difference calculated at each voxel at the step (b), by converting the phase difference at each voxel calculated at the step (b) into a corresponding temperature; and

CLPV:
(d) displaying the temperature distribution image constructed at the step (c).

CLPV:
(e) detecting a displacement of the target body between a collection of the chemical shift data with the temperature change and a collection of the chemical shift data without the temperature change according to a phase change in the chemical shift data collected from the temperature independent portion of the imaging target region with and without the temperature change.

CLPV:
(f) adjusting strengths of gradient magnetic fields used by the nuclear magnetic resonance imaging unit at the step (a), when the displacement is detected at the step (e), to shift the imaging target region sequentially to a position at which the phase change is nullified.

ORPL:
Vidrine et al, "Feedback Excitation Nuclear Magnetic Resonance Spectrometry and its Application to Simultaneous Temperature Measurement", Analytical Chemistry, vol. 50:293-303, (1978).

ORPL:
Hall et al., "Mapping of pH and Temperature Distribution Using Chemical-Shift-Resolved Tomography", Journal of Magnetic Resonance, vol. 65:501-505, (1985).

ORPL:
Lebihan et al, "Non-Invasive Temperature Mapping During Hyperthermia by MR Imaging of Molecular Diffusion", Magnetic Resonance Imaging II, pp.342-343.

Generate Collection

L10: Entry 2 of 3 File: USPT Apr 25, 2000

DOCUMENT-IDENTIFIER: US 6054855 A
TITLE: Magnetic susceptibility control of superconducting materials in nuclear magnetic resonance (NMR) probes

ABPL:
A method and apparatus which utilizes the hysteritic behavior of type II superconductors for reducing the effective magnetic susceptibility of high temperature superconducting materials used close to the sample region in nuclear magnetic system probes. The method is particularly applicable to receiver coils. Reducing the magnetic susceptibility of superconducting receiver coils enables the improved sensitivity they inherently provide to be realized without loss of resolution resulting from line broadening caused by susceptibility discontinuities of materials near the sample region of the probe.

BSPR:
This invention relates to the field of nuclear magnetic resonance apparatus and in particular to probe structures incorporating high temperature superconductor (HTS) materials, and more particularly to methods and apparatus for minimizing perturbations of the polarizing and rf magnetic fields caused by HTS materials present in the region of the sample.

BSPR:
Nuclear magnetic resonance (NMR) spectrometers first became available in 1946. In 1950 observations of "shifted" resonance in nitrogen spectra by W. G. Proctor & F. C. Yu, Phys. Rev. 77, 717, (1950) stimulated efforts to improve the homogeneity and stability of magnets used in the experiments and led to the observation of chemically shifted resonances in proton spectra by J. T. Arnold, S. S. Dharmatti, and M. E. Packard, Jour. Chem. Phys. 19, 1608, (1951). This marked the beginning of high resolution NMR and its application as an analytical tool for chemistry, and sparked rapid growth in the development of NMR spectrometers. This development continues today at a pace limited only by the availability of relevant technology. The present work is predicated upon improvements in rf probe performance incorporating receiver coils and other parts made from recently available high temperature superconducting (HTS) materials. HTS materials are type II superconductors. The terms "HTS materials" and "type II superconductors" will hereafter be used interchangeably herein.

BSPR:
Nuclei of most isotopes of the elements have non-zero spin and exhibit gyromagnetic properties. These non-zero spin nuclei behave like microscopic spinning bar magnets. When a static homogeneous magnetic field B is applied to an ensemble of spin active nuclei, the spins align, some in the direction of the field and some in the direction opposed to the field. A net polarization of the ensemble of spins in the direction of the field results and the spins are said to be "polarized" by the field. If a polarized ensemble of nuclei is simultaneously subjected to an rf magnetic field, usually called the B.sub.1 field, said B.sub.1 field having an appropriate frequency and spatial orientation with respect to the polarizing field B, an NMR response signal can be generated.

BSPR:
The broad general utility of NMR as a tool for determining the chemical structure of compounds is due to the influence of the molecular environment on the local magnetic field at the nuclei. The local magnetic field at the nucleus of a particular nuclear species at a particular site in a molecule is the vector addition of the externally applied field and the field caused by the magnetic influence of its molecular environment. By way of example, circulation of electrons about the nucleus caused by the applied field results in an induced field at the nucleus which in some instances opposes the applied field (diamagnetism), and in some instances augments it (paramagnetism). By way of further example the local field at a nucleus can be additionally modified, taking on multiple values or "splitting" due to interactions with other spin active nuclei in the molecule. These two effects, known as "chemical shift" and "spin-spin coupling" respectively, are major sources of the fine structure

seen in NMR spectra as mo[redacted] fully described in "Introducti[redacted]To NMR Spectroscopy", R. Abrahms, J Fisher, P. Loftus, J. Wiley & Sons, 1993, chap. [redacted] pp. 13-33, chap. 3, pp. 34-59. NMR spectra which are characterized by resonance lines that are narrower than the shifts in resonance caused by chemical shift and spin-spin coupling are known as high resolution spectra. These lines are primarily made possible to observe by the application of an extremely homogeneous polarizing field. The frequency of the NMR response signal is proportional to the local magnetic field at the nuclei, the proportionality constant being .gamma., the magnetogyric ratio. Any slight deviation from homogeneity of the local magnetic field over the sample region causes a corresponding shift in the resonance of the nuclei affected resulting in undesirable line broadening of the response signal.

BSPR:
An NMR spectrometer is comprised of: 1) a DC magnet which provides the stable, homogeneous, static magnetic field required for polarizing the spins, 2) an rf system which provides a suitable rf excitation signal, 3) a coil or a plurality of coils for coupling the rf excitation to the spins and for receiving the NMR response signal, 4) a detection system for detecting the NMR response signal, 5) a signal processing system for processing the detected NMR response signal, and 6) an output device for displaying the NoM response signal. For high resolution NMR studies, the compound under investigation is usually dissolved in or mixed with a suitable solvent and is in liquid form contained in a sample tube which is typically 5 mm in diameter. The apparatus known as the probe holds the sample in a sample holder portion of a probe in the most homogeneous region in the magnetic field. The coil or coils for coupling the rf excitation to the sample and for detecting the NMR response signal are also mounted to the probe.

BSPR:
NMR is an inherently insensitive technique. Sensitivity is strictly defined in terms of the minimum concentration of a test material required to produce a signal that is just detectable above the level of noise. For practical purposes however, the signal to noise ratio, S/N, is generally considered a good measure of sensitivity. Continued improvement in sensitivity has been a constant objective in the development of NMR spectrometers. Increasing signal strength, reducing noise, and improving signal processing methods have all contributed to this improvement. Many of the factors that influence the attainable signal to noise ratio are treated in "A Handbook of Nuclear Magnetic Resonance", R. Freeman, Longman Scientific & Technical, 1988, pp. 216-229 which is hereby incorporated herein by reference.

BSPR:
In addition to sensitivity, resolution of spectral information is an important aspect of NMR spectrometer performance. Natural line widths can be narrow for liquid samples, less than 1 Hz by way of example. To avoid degrading the resolution, both the static magnetic field B and the rf excitation field B.sub.1 should be homogeneous over the volume of the sample, and stable over the time of data acquisition to the order of 1 part in 10.sup.9. The data acquisition time can be very long, particularly when acquiring the spectra of nuclei other than protons, such as .sup.13 C by way of example. For .sup.13 C nuclei using natural abundance samples, the overall sensitivity relative to .sup.1 H is 1.7.times.10.sup.-4. The direct observation of .sup.13 C nuclei therefore typically requires many scans and may require averaging the NMR responses over hours or days to achieve the required signal to noise ratio. Any small change in the magnetic field over this time period will cause the NMR signal to shift slightly and effectively broaden the resonance response. Field homogeneity requirements are addressed by careful magnet design, the use of shimming coils and by spinning the sample. Field-frequency lock systems, such as described in "Modern NMR Spectroscopy", J. K. M. Sanders & B. K. Hunter, Oxford University Press, 1993, chap. 1, pp. 39-41. are used to achieve the required stability.

BSPR:
The probe is a critical component in an NMR spectrometer. For a given static magnetic field strength and a given sample size, the performance of the probe largely determines the sensitivity of the spectrometer. An important consideration in probe design is the coupling efficiency .zeta. of the receiver coil to the sample. .zeta. is the ratio of effective inductance to total inductance of the receiver coil. Any portion of the inductance of the receiver coil that does not contribute towards the detection of the NMR signal, such as the inductance of the leads of the receiver coil by way of example, results in a loss of sensitivity proportional to .zeta..sup.1/2. Another important consideration is the quality factor Q of the receiver coil which affects sensitivity by a factor of Q.sup.1/2, since signal voltage is proportional to Q and noise voltage is proportional to Q.sup.1/2. Q represents the ratio of energy stored in the receiver coil resonant circuit to the energy dissipated through resistive losses in the circuit. Another important consideration in probe design is the receiver-coil filling factor .xi. which, for a fixed coil volume, influences the signal strength and the sensitivity directly. .xi. is a measure of the energy stored

in the transverse magnet[redacted]field coupling to the sample, c[redacted]ared to the total magnetic energy stored i[redacted]he receiver coil resonant circu[redacted] Filling factor .xi., coupling efficiency .zeta., and quality factor Q should all be as large as possible for maximum sensitivity.

BSPR:
Modern spectrometers use superconducting DC magnets for producing the static polarizing field. The sample is placed in a cylindrical tube positioned coaxial with the DC magnet. Transmitter and receiver coils made of normal, i.e. non-superconducting, materials can be saddle coils as shown in FIG. 1a or split formed-wire coils as shown in FIG. 1b. Either are ordinarily shaped to couple closely to the sample while providing the radio frequency B.sub.1 field orthogonal to the static field. Coils made of high temperature superconducting (HTS) films are very attractive for use in NMR spectrometers because of their low rf resistance and resulting low noise. Using HTS materials, coils have been fabricated by depositing a thin layer of superconductor on a flat substrate. A pair of such coils forming a magnetically coupled system known as a Helmholtz pair, placed on opposite sides of a sample, is shown in FIG. 2a. A second pair of similar HTS coils can be positioned orthogonal to the first pair as shown in FIG. 2b to provide for a field-frequency lock signal.

BSPR:
Best results are obtained with HTS coils when the superconductor is lattice matched to the substrate, i.e. grown epitaxially. The substrate should be a thermally conductive material to facilitate cooling of the coil and should have low magnetic susceptibility to avoid degrading the homogeneity of the magnetic field. Acceptable substrate materials include sapphire, lanthanum aluminate, and magnesium oxide. A preferred HTS material is YBa.sub.2 Cu.sub.3 O.sub.7-.delta. (YBCO), which has a critical transition temperature T.sub.C of approximately 87.degree. K. A coil made of this material is described in "HTS Receiver Coils For Magnetic Resonance Instruments", R. S. Withers, B. F. Cole, M. E. Johansson, G. C. Laing, G. Zaharchuk, Proc. SPIE, 2156, 27-35, (1994). Another Class II superconductive material useful in this coil application is Tl.sub.2 Ba.sub.2 CaCu.sub.2 O.sub.8.

BSPR:
For proper performance HTS coils must be maintained at a temperature significantly below their superconducting transition temperature T.sub.C. U.S. Pat. No. 5,508,613 entitled Apparatus For Cooling AMR Coils, to Vincent Kotsubo and Robert D. Black, describes an apparatus for cooling HTS coils as required for proper operation. A particular embodiment incorporates a Joule-Thomson or Cillord-McMahon closed cycle refrigeration unit which cools the coils to 25.degree. K. The coils are generally thermally isolated from the samples in this apparatus and the samples can be maintained at or near room temperature if desired.

BSPR:
High resolution NMR probes using HTS coils can provide higher sensitivity than probes with non-superconducting coils. For a given sample volume the sensitivity of a coil is proportional to (.xi.Q/T).sup.1/2, where T is the coil temperature and .xi. and Q are the aforementioned filling factor and quality factor respectively. A superconducting coil may have a Q of 20,000 as compared with a Q of 250 for a room temperature coil and is typically operated at 25.degree. K. as compared with 300.degree. K. for a room temperature coil. With the geometry appropriate for a 5 mm. sample tube, and allowing for the loss of filling factor required for thermal isolation of the sample from the coil, the potential sensitivity gain can approach a factor of 10.

BSPR:
It is known in the art that the probe materials and sample materials can cause significant distortion of the polarizing and rf magnetic fields due to their susceptibility. To achieve high resolution spectra, these distortions must be controlled and/or corrected. In particular, abrupt changes in susceptibility near the sensitive sample region of the probe can cause serious degradation of the field uniformity at the sample region, which can generally be partially corrected with shim coils. The aforementioned field distortion can be minimized by using cylindrical symmetric components and positioning material boundaries as far removed a possible from the sample region. Additionally, careful choice of the materials used in the probe is of paramount importance. Materials normally used in NMR probes have diamagnetic volume susceptibilities of several parts per million.

BSPR:
The best known characteristic of superconductors is their ability to carry a steady current without any power loss., i.e., without any associated voltage drop. Complete magnetic flux expulsion, commonly known as the Meissner effect, is a second fundamental characteristic of superconductivity. The class of superconducting

materials which complete[redacted]expel flux from their bulk volu[redacted] thereby maintaining a condition of zero flux d[redacted]ity internally, are known as ty[redacted] I superconductors. A type I superconductor is perfectly diamagnetic. Type I superconductors are characterized by a low critical transition temperature T.sub.C and a single critical magnetic field H.sub.C (T) with a relatively small range.

BSPR:
A large class of materials known as type II superconductors allow flux to enter the bulk of their volume in a special way and in small quantized amounts, as described in "Foundations of Applied Superconductivity", Orlando and Delin, Addison Wesley Publishing Co., 1990, chap's 6, 7, pp. 259-391, which is hereby incorporated by reference. Type II superconductors typically have a higher critical transition temperature T.sub.C than Type I superconductors and they have two critical fields, H.sub.C1 (T) and H.sub.C2 (T). For values of H temperature range. Type II superconductors therefore are practical for, and useful in, engineering applications such as NMR probe coils.

BSPR:
The relationship between induced magnetization M and applied field H is much more complicated for a type II superconductor than for a type I superconductor. As heretofore mentioned, flux vortices penetrate into the bulk volume of the superconductor at applied fields H.gtoreq.H.sub.C1. The type II superconducting material is constituted to provide pinning forces for the purpose of inhibiting lateral movement of the vortices when an externally driven current is passed through the material. Such vortex movement would cause undesired power losses. Because of the pinning forces however, the flux vortices, after penetrating the surface when the applied field exceeds H.sub.C1, are not uniformly distributed throughout the bulk of the superconductor in an equilibrium lattice, but instead are bunched up near the surface. As the applied field is further increased above H.sub.C1 the flux vortices are forced further into the superconductor but they remain non uniformly distributed throughout the bulk volume.

BSPR:
According to the critical state model, as an external magnetic field is applied to the superconducting material, surface currents are set up that flow in such a direction as to exclude magnetic flux from the interior of the material. However there is a limiting current density J.sub.C (H) that the superconductor can carry. The model assumes that there are only three states of current flow possible with a given magnetic field axis, one being zero current density for regions that have never felt the magnetic field. The other two are full current flow J.sub.C (H) perpendicular to the axis, but each are of opposite sense from the other depending on the sense of the electromotive force that accompanied the last local change of applied field. These, local currents contribute to the magnetization of the material and thereby influence its effective susceptibility. FIGS. 6a, 6c and 6e show the locally averaged magnetic flux density distributions in a thin film superconductor of thickness 2a for different values of an increasing applied magnetic field H. The field is oriented parallel to the surface of the superconductor. FIGS. 6b, 6d and 6f show the corresponding current density profiles. The applied field H at which the flux fully penetrates the film is known as the penetration field, which will hereafter be designated Hp herein. It can be shown for the aforementioned thin film of thickness 2a, that Hp=J.sub.C (H)(a) and that the thermodynamic magnetization is -Hp/2. At field values of Hp and above the effective susceptibility ##EQU2## is equal to -0.5 J.sub.C (H) a/H.

BSPR:
To maintain homogeneity of the magnetic fields in the sample region of an NMR spectrometer, the most critical probe component which must be considered is the coil, because it is generally closest to the sample region and inevitably includes some susceptibility discontinuities in its geometry. For normal coil materials, i.e. non-superconducting, coil field perturbations can be minimized by constructing the coils of materials such that the overall coil structure exhibits a low average value of magnetic susceptibility. This is accomplished by making the coils from a composite material with diamagnetic and paramagnetic components using methods such as electroplating by way of example, to produce a sandwich structure of the two types of materials. Overall high electrical conductivity is maintained for this structure. Suitable diamagnetic materials include copper, silver and gold. Suitable paramagnetic materials include aluminum, rhodium and platinum.

BSPR:
However, when employing HTS materials as probe coils to realize the aforementioned higher sensitivity, the use of sandwich structures of the two types of susceptibility materials as described in connection with normal materials above is not an available option. Therefore the improved sensitivity under these circumstances has generally been achievable only at the cost of degraded resolution. The tradeoff of degraded spectral resolution for improved sensitivity has heretofore limited realization of the full potential inherent in the use of HTS materials in NMR probes.

BSPR:
We have provided a method and apparatus for reducing the effective magnetic susceptibility of HTS materials used in NMR probes to a near zero value. This significantly reduces distortions of the homogeneous polarizing and rf magnetic fields, said distortions being caused by discontinuities in susceptibility at material boundaries in the probe. Broadening of spectral lines and consequent degradation of spectral resolution resulting from distortion of the magnetic fields can thereby be minimized or even eliminated. The improved sensitivity inherently achievable using a HTS receiver coil can be fully realized using this invention without accompanying degradation of spectral resolution resulting from magnetic field distortion.

BSPR:
An object of this invention is to provide a high sensitivity, high resolution NMR probe.

BSPR:
Another object of this invention is to control the magnetic susceptibility of type II superconducting materials used in NMR probes.

BSPR:
Another object of this invention is to minimize distortions of the magnetic fields caused by type II superconducting materials used in NMR probes.

BSPR:
Another object of this invention is to minimize distortions of the magnetic fields in the sample region of NMR probes.

BSPR:
Another object of this invention is to provide a method for demagnetizing type II superconducting materials used in NMR probes.

BSPR:
Another object of this invention is to provide a method for optimally demagnetizing rf probe coils in NMR probes.

DRPR:
FIG. 6b shows, for Hcurrent density profile in a thin film.

DRPR:
FIG. 6d shows, for H=H.sub.p, a locally averaged current density profile in a thin film.

DRPR:
FIG. 6f shows, for H>H.sub.p, a locally averaged current density profile in a thin film.

DRPR:
FIG. 10 shows plots of the magnetization and current density in a superconducting ilm after a single pulse demagnetization.

DRPR:
FIG. 12 shows a plot of current density in a superconducting film after an AC demagnetization procedure.

DEPR:
With reference to FIG. 2a, a pair of prior-art planar thin film so called high temperature superconductors (HTS) probe coils 5, 5' forming a Helmholtz coil pair 6 is shown schematically disposed on opposite sides of a cylindrical sample 8. The substrates on which the coils 5, 5' are deposited are assumed but not shown. In FIG. 2b, a second prior-art Helmholtz coil pair 10 is shown disposed orthogonal to the coil pair 6. Substrates for coil pair 10 are also assumed but not shown. In FIG. 2c, a sectional view through the coil pair 6 and the cylindrical sample 8 of FIG. 1a is shown, including the substrate 12 on which coils 5, 5' are deposited. Such HTS coils significantly disturb the homogeneity of the magnetic field in the sample region of the probe.

DEPR:
With reference to FIG. 4a to FIG. 4c, the Meissner effect magnetic field distribution within a perfectly uniform diamagnetic conductor 34 and the region proximate to it 36 are represented as the superposition of 1) the uniform applied magnetic field 38 in the absence of the conductor and 2) the field due to the induced magnetization M 40 which represents the inherent magnetic property of the material. A corresponding representation can be made for a uniformly magnetized paramagnetic material. Inside the conductor, the magnetic flux density B=.mu..sub.0 (H+M) where .mu..sub.0 is the permeability of free space, H is the applied magnetic field and M is the induced magnetization. The relationship M=.sub..chi.m H defines the magnetic susceptibility .sub..chi.m which is negative for diamagnetic materials and positive for paramagnetic materials. As discussed, within a type I superconductor, which is a perfectly diamagnetic medium, B=0. Therefore, M=-H and the magnetic susceptibility .sub..chi.m =-1. Because type I superconductors have critical transition temperatures which are very low, i.e. 4 degrees Kelvin, and limited magnetic field range, they have not been generally useful in NMR probe engineering applications.

DEPR:
For the case where the applied magnetic field is parallel to the surface of a type II superconductor, FIGS. 6a through 6f are prior known illustrations showing magnetic flux density distributions and the corresponding current density profiles within such a thin film Class II superconductor.

DEPR:
With reference to FIG. 6a, the locally averaged flux density distribution 52 within the thin film having a thickness 2a, as a function of position in the film, is shown for the applied field 54 Hcurrent density J.sub.C profile 56 and shows that the current flow is limited to the depth of penetration 58 of the field. FIG. 6c shows the flux density distribution 60 for H=Hp at which there is full penetration of the field into the superconductor. FIG. 6d shows the corresponding uniform current density J.sub.C 62 throughout the bulk volume of the superconductor. FIG. 6e shows the flux density distribution 63 for H>H.sub.p. The current density 62 as shown in FIG. 6f remains uniform throughout the superconductor at its maximum value J.sub.C.

DEPR:
With reference to FIG. 8, for the aforementioned case of FIG. 7b, i.e. applied field 68 parallel to surface 71 of coil 64, the thermodynamic fields 44, /.mu..sub.0 72, and 50 are shown as functions of the applied field H 68 as H is increased from zero. The thermodynamic magnetization remains constant as the applied field H 68 is increased above the penetration field H.sub.p 69 because the current distribution for the locally averaged current density J remains the same.

DEPR:
The inventive method for demagnetizing and reducing the effective susceptibility is described in connection with FIGS. 9a through 12. With reference to FIG. 9a, a first embodiment of the inventive method is shown

for bringing the magnetization of the HTS thin film receiver coil to zero or near zero. Point A, 74 is the state of magnetization of the film after being placed in the axial field H.sub.0 76 of the polarizing magnet. The magnetic flux density B corresponding H.sub.0 76 is typically between 4.7 Tesla and 23 Tesla for spectrometer applications of interest. At point A, 74 the film exhibits magnetization which may cause broadening of spectral lines and be detrimental to the spectrometer resolution. If the axial magnetic field H 68 can be increased by .DELTA.H, 71 to H.sub.0 +.DELTA.H 79, the magnetization could advance to point C, 78. One way this can be accomplished is by applying a brief pulse of current to a demagnetizing coil 1 surrounding the NMR probe as shown in FIG. 13. When the current in the pulse is reduced to zero bringing H back to its original value H.sub.0 76, the magnetization in the film moves along the hysteresis path 80 towards zero at point D, 81. can be shown that the value of .DELTA.H required to bring the magnetization 50 to zero is approximately equal to the product of the HTS film's critical current density J.sub.C and its thickness 2a. i.e. .DELTA.H=(2-.sqroot.2)a J.sub.C.

DEPR:
As shown in FIG. 9b, the sign of the magnetization 50 can be reversed if an excessively large current pulse is applied and then removed from the demagnetizing coil. The magnetization 50 will follow the path from point E, 82 to point F, 83 and then to point G, 84. This magnetization would also be disturbing to the homogeneity. However, by then applying a demagnetizing current in the reverse direction in the demagnetization coil 1, the magnetization can be brought back to zero via the path 86 from point G, 84 to point K, 88. When the pulse stops, the magnetization will trace the path 90 from point K, 88 to point L, 92. At point L, 92 the field is back to its initial value H.sub.0 and the magnetization and the effective susceptibility ##EQU3## are zero or near zero.

DEPR:
With reference to FIG. 10, profiles of the magnetization M 94 and the current density J 96 in the superconducting thin film as a function of position are shown. This is the profile after the field has been brought back to its initial value H.sub.0 as shown in the process of FIG. 9a. After the pulse stops, as the magnetic field is decreasing back towards the initial value H.sub.0, the induced currents 98, 100 in the outer regions of the superconducting material reverse their directions and the thermodynamic magnetization and the effective susceptibility ##EQU4## are reduced to zero or near zero when the area from x=a to x=-a inder curve 94 sums to zero.

DEPR:
In the embodiment of the invention described in FIGS. 9a and 10 only a single brief application of current is applied to the demagnetizing coil to achieve reduced magnetization. However in this single pulse embodiment, accurate knowledge of the values of J.sub.C (H.sub.0) and the film thickness 2.sub.a are required. FIG. 9b illustrates an extension of the invention described in FIGS. 9a and 10. The reversed brief pulse of current to the demagnetizing coil, which cause the change in magnetization from point G, 84 to point K, 88 in FIG. 9b corrects for an excessively large initial pulse. By providing several alternately positive and negative brief pulses of current to the demagnetizing coil a more complete demagnetization of the probe coils is achievable.

DEPR:
With reference to FIG. 12, the current density distribution 118 as a function of position in the film is shown which corresponds to a linear decaying demagnetization drive. Instead of the current quadrupole as shown in FIG. 10 that results from the single pulse technique, the AC drive technique produces a current multipole of higher order resulting in even lower stray magnetic fields.

DEPR:
FIG. 13 shows an NMR system including a demagnetization coil 1 shown interposed between a HTS thin film coil pair 6 and the main field winding coil 124. The demagnetization coil 1 is connected electrically to the control/power supply 150 which is coupled to a computer 153 via a bus 154. The computer includes memory. The structure of the remainder of the NMR system is standard. Sample holder 10 is shown

schematically axially positioned within a space where very high magnetic fields are provided by main field coils 124 under DC supply control 126. The HTS coil pair 6 is mounted to a substrate 12 which is held in a heat transfer base for cooling via gas flow from cryostat 127 through conduits 125 and 129. The HTS probe coils are shown in the vacuum vessel 155 to reduce heat transfer. The demagnetization coil is also shown inside the vacuum vessel but since this is a normal coil it does not need to be inside the vessel and would be mounted to the outside of the vessel. The probe coil pair is coupled to the RF transmitter 132 and receiver 134 via transmit/receive switch 130 and loop antenna 156. Coil 1 can also be a plurality of coils.

DEPR:
The AC demagnetization technique is the subject of copending patent application Ser. # U.S. Pat. No. 5,986,453, entitled "An AC Magnetic Susceptibility Control of Superconducting Materials in Nuclear Magnetic Resonance (NMR) Probes", of which I am a joint inventor and which is filed concurrently herewith.

DEPR:
In the above description of the invention the applied field H has been assumed to be parallel to the face of the superconductive film, and the critical currents J.sub.C to flow in one direction along the +y-axis on one side of the superconductive film and in the opposite direction along the -y-axis on the other side of the film. In many cases the surface of the fill may not be perfectly aligned with field or may be normal or nearly normal to the direction of the applied field. In this case critical current attempts to flow in planes normal to the applied field components, constrained of course to the boundaries of the subconductive film. Since the width of the superconductive coil structure (71' in FIG. 7b) can be and normally is large compared to the thickness of the superconducting film (71 in FIG. 7b), the effect of these other currents is to form larger loops and corresponding greater degree of magnetic field distortion over the sample volume. In this case the demagnetization can be carried out by also applying a demagnetizing field normal to the surface of the superconductive film. The same techniques of single pulsing, multiple pulsing with pulses of opposite field direction, or applying a slowly decaying AC field may be used.

DEPR:
In the case of a complex superconducting probe coil geometry it may be desirable to have an array of demagnetizing coils. Each coil could control the maximum field excursion seen by a different part of the superconducting rf probe coil structure. This arrangement might provide better overall control of the compensation. By way of example one demagnetizing coil could provide an approximately uniform field over the entire probe coil structure while another could be in the form of a linear gradient demagnetizing coil that could provide greater demagnetizing fields at the ends of the structure.

DEPR:
The demagnetization process must be carried out each time a probe containing an rf probe coil is inserted into the magnet. Each different rf probe coil may require a different recipe for demagnetization, which could be preserved in a data bank. Each time a probe is inserted into the magnet the demagnetizing coil power supply could be programmed to read the identifying data for the particular probe and automatically provide the correct demagnetizing process.

DEPR:
An essential feature of the inventive process described herein is utilization of the hysteritic behavior of the type II superconductor material. Because of this hysteritic behavior, the thermodynamic magnetization of superconducting components used in NMR probes may be reduced to zero or near zero when, after insertion into the polarizing field of the spectrometer, they are temporarily exposed to additional appropriate demagnetizing fields.

DEPR:
Although demagnetizing fields described herein are attributed to currents passed through demagnetizing coils, said coils surrounding the superconducting components, it is not intended that this invention be

restricted to demagnetizing fie[redacted]produced in that manner. Rather, it is i[redacted]ded that the invention be interpreted broadly as being applicable to demagnetizing fields produced in any manner whatsoever. By way of example of an alternative to demagnetizing coils, a suitable incremental demagnetizing field parallel to the surface of a thin film superconducting coil may be briefly provided by a transient decaying oscillatory movement of the superconducting coil relative to the main static polarizing field. An embodiment utilizing this alternative is the subject of my copending patent application, Ser. No. 08/965,899, entitled "Nuclear Magnetic Resonance Methods and apparatus", which is filed concurrently herewith. Although the demagnetizing process described herein is applied to superconducting NMR receiver coils, it is not intended that the inventive process be so restricted. Rather it is intended that the process be applicable to any superconducting component part used in NMR probes when homogeneity is essential such as Faraday shields by way of example. In accordance with these considerations, the scope of the invention should be construed in view of my claims. With this in mind,

CLPR:
1. An NMR system comprising:

CLPR:
2. The NMR system of claim 1 further including;

CLPR:
3. A nuclear magnetic resonance (NMR) system comprising,

CLPR:
4. The NMR system of claim 3, wherein said apparatus for changing the magnitude of said magnetic field comprises,

CLPR:
5. The NMR system of claim 4 further comprising a computer and a data bank,

CLPR:
6. The NMR system of claim 5, wherein said current supplied to said demagnetizing coils is a single brief pulse of current which induces an incremental change in the magnetic field.

CLPR:
7. The NMR system of claim 6, wherein the magnitude of the incremental change in said magnetic field in the region of said superconducting probe coils caused by said single brief pulse of current is equal to {2-.sqroot.2} times the product of the critical current density J.sub.C of said superconducting coils and the thickness of said superconducting coils, resulting, on termination of said brief pulse of current, in said magnetization of said superconducting probe coils traversing a hysteritic path to zero or near zero in the magnetization verses applied-field space.

CLPR:
8. The NMR system of claim 7, wherein said superconducting probe coil material is YBCO family.

CLPR:
9. The NMR system of claim 5, wherein said controller is capable of supplying current to said demagnetizing coils comprised of two consecutively applied brief pulses of current.

CLPR:
10. The NMR system of claim 9, wherein said magnitude of the incremental change in said magnetic field in the region of said superconducting probe coils caused by said first single brief pulse of current is greater than the {2-.sqroot.2} times the product of the critical current density J.sub.C of said superconducting coils and the thickness of said superconducting coils, resulting, on termination of said first brief pulse of current,

in said magnetization of said superconducting probe coils being reversed in sign as said magnetization traverses a hysteritic path beyond zero in the magnetization verses applied-field space, and wherein said current controller is capable of providing reverse polarity current pulses so that the second of said consecutively applied brief pulses of current is reversed in sense from said first of said consecutively applied brief pulses of current whereby on termination of said second brief pulse of current, said magnetization of said superconducting probe coils traverses a hysteritic path to zero or near zero in the magnetization verses applied-field space.

CLPR:
11. The NMR system of claim 10, wherein said superconducting probe coil material is selected from the group consisting of YBa.sub.2 Cu.sub.3 O.sub.7-.delta. (YBCO) and Tl.sub.2 Ba.sub.2 CaCu.sub.2 O.sub.8.

CLPR:
12. The NMR system of claim 5, wherein said current supplied to said demagnetizing coils is comprised of two or more consecutively applied brief pulses of current, each of said consecutively applied brief pulses of current being of a sense and magnitude to cause said magnetization of said superconducting probe coil to approach closer to zero.

CLPR:
13. The NMR system of claim 12, wherein said superconducting probe coil material is selected from the group consisting of YBa.sub.2 Cu.sub.3 O.sub.7-.delta. (YBCO) Tl.sub.2 Ba.sub.2 CaCu.sub.2 O.sub.8.

CLPR:
14. The NMR system of claim 4, wherein said current supplied to said demagnetizing coils is a single brief pulse of current.

CLPR:
15. The NMR system of claim 14, wherein said current is adjusted so that the magnitude of the incremental change in said magnetic field in the region of said superconducting probe coils caused by said single brief pulse of current is equal to {2-.sqroot.2} times the product of the critical current density J.sub.C of said superconducting coils and 2) the thickness of said superconducting coils, whereby on termination of said brief pulse of current, said magnetization of said superconducting probe coils traverses a hysteritic path to zero or near zero in the magnetization verses applied-field space.

CLPR:
16. The NMR system of claim 15, wherein said superconducting probe coil material is YBa.sub.2 Cu.sub.3 O.sub.7-.delta. (YBCO).

CLPR:
17. The NMR system of claim 4 wherein said current supplied to said demagnetizing coil is comprised of two consecutively applied brief pulses of current.

CLPR:
18. The NMR system of claim 17, wherein said current is adjusted so that the magnitude of the incremental change in said magnetic field in the region of said superconducting probe coils caused by said first single brief pulse of current is greater than the product of 1) the critical current density J.sub.C of said superconducting coils and 2) the thickness of said superconducting coils, resulting, on termination of said first brief pulse of current, in said magnetization of said superconducting probe coils being reversed in sign as said magnetization traverses a hysteritic path beyond zero in the magnetization verses applied-field space, and wherein the second of said consecutively applied brief pulses of current is reversed in sense from said first of said consecutively applied brief pulses of current, resulting, on termination of said second brief pulse of current, in said magnetization of said superconducting probe coils traversing a hysteritic path to

zero or near zero in the magnetization verses applied-field space.

CLPR:
19. The NMR system of claim 18, wherein said superconducting probe coil material is selected from the group consisting of YBa.sub.2 Cu.sub.3 O.sub.7-.delta. (YBCO) and Tl.sub.2 Ba.sub.2 CaCu.sub.2 O.sub.8.

CLPR:
20. The NMR system of claim 4, wherein said current supplied to said demagnetizing coils is comprised of two or more consecutively applied brief pulses of current, each of said consecutively applied brief pulses of current being of a sense and magnitude to cause the magnetization of said superconducting probe coil to approach closer to zero.

CLPR:
21. The NMR system of claim 20, wherein said superconducting probe coil material is selected from the group consisting of YBa.sub.2 Cu.sub.3 O.sub.7-.delta. (YBCO) and Tl.sub.2 Ba.sub.2 CaCu.sub.2 O.sub.8.

CLPR:
22. In a method for improving the homogeneity of the magnetic field in a sample region of an NMR spectrometer, said spectrometer comprising a probe having a sample region, a high temperature superconducting coil pair proximate to said sample region for exciting resonance in, and detecting NMR responses from a sample, and a magnet for providing a homogeneous static polarizing field, the improvement comprising:

CLPR:
23. A process for demagnetizing superconducting rf probe coils used in nuclear magnetic resonance (NMR) systems where said NMR system includes a magnet to produce a static polarizing magnetic field, a probe for inserting into said magnetic field, said probe having a sample region therein, said probe containing one or more rf probe coils proximate to said sample region to provide magnetic coupling to any sample material in said sample region, at least one of said rf probe coils being composed of superconducting material, means for cooling said superconducting probe coils, an rf transmitter for supplying rf energy to at least one of said rf probe coils, an rf receiver coupled to said superconducting probe coil, said receiver amplifying and detecting any signal from said sample material, the process including

CLPR:
24. The process of claim 23 wherein said step of changing the magnitude of magnetic field lines pervading to said superconductor coil pairs is accomplished by supplying current to demagnetization coils configured to change the field parallel to said superconductor coil pair.

CLPR:
26. The process of claim 25, wherein said current supplied to said demagnetizing coils is a single brief pulse of current.

CLPR:
27. The process of claim 26, wherein the magnitude of the change in said magnetic field in the region of said superconducting probe coils caused by said single brief pulse of current is equal to the product of 1) the critical current density J.sub.C of said superconducting coils and 2) the thickness of said superconducting coils, resulting, on termination of said brief pulse of current, in said magnetization of said superconducting probe coils traversing a hysteritic path to zero or near zero in the magnetization verses applied-field space.

CLPR:
29. The process of claim 25, wherein said current supplied to said demagnetizing coils is comprised of two consecutively applied brief pulses of current.

CLPR:
30. The process of claim 29, wherein the magnitude of the change in said magnetic field in the region of said superconducting probe coils caused by said first single brief pulse of current is greater than the product of 1) the critical current density J.sub.C of said superconducting coils and 2) the thickness of said superconducting coils, resulting, on termination of said first brief pulse of current, in said magnetization of said superconducting probe coils being reversed in sign as said magnetization traverses a hysteritic path beyond zero in the magnetization verses applied-field space, and wherein the second of said consecutively applied brief pulses of current is reversed in sense from said first of said consecutively applied brief pulses of current, resulting, on termination of said second brief pulse of current, in said magnetization of said superconducting probe coils traversing a hysteritic path to zero or near zero in the magnetization verses applied-field space.

CLPR:
32. The process of claim 25 wherein said current supplied to said demagnetizing coils is comprised of two or more consecutively applied brief pulses of current, each of said consecutively applied brief pulses of current being of a sense and magnitude to cause said magnetization of said superconducting probe coil to approach closer to zero.

CLPR:
34. The process of claim 23 wherein the step of changing the magnetic field lines pervading said superconductor coil pair is accomplished by supplying current to demagnetization coils configured to provide magnetic field lines perpendicular to said superconducting coil pair.

CLPR:
36. The process of claim 35, wherein said current supplied to said demagnetizing coils is a single brief pulse of current.

CLPR:
37. The process of claim 35, wherein said current supplied to said demagnetizing coils is comprised of two consecutively applied brief pulses of current.

CLPR:
38. The process of claim 35, wherein the magnitude of the change in said magnetic field it the region of said superconducting probe coils caused by said first single brief pulse of current is greater than the product of 1) the critical current density J.sub.C of said superconducting coils and 2) the thickness of said superconducting coils, resulting, on termination of said first brief pulse of current, in said magnetization of said superconducting probe coils being reversed in sign as said magnetization traverses a hysteritic path beyond zero in the magnetization verses applied-field space, and wherein the second of said consecutively applied brief pulses of current is reversed in sense from said first of said consecutively applied brief pulses of current resulting, on termination of said second brief pulse of current, in said magnetization of said superconducting probe coils traversing a hysteritic path to zero or near zero in the magnetization verses applied-field space.

CLPV:
a probe, a coil in said probe, said coil being a superconducting coil pair made from class II high temperature superconducting materials, said pair being thin film coils;

CLPV:
a current controller for controlling the current from said demagnetization power supply to said

demagnetization coil.

CLPV:
one or more power supplies for supplying current to said demagnetizing coils; and

CLPV:
a current controller to control the current from said power supplies.

CLPV:
said probe having computer readable identification marking thereupon such that on insertion into said NMR system said computer is able to identify said probe and choose from said data bank appropriate data comprised of a recipe for demagnetizing said superconducting coils in said probe;

CLPV:
said recipe enabling said current controller to provide the correct current from said power supply to said demagnetizing coils for said probe such that the magnetization of aid superconducting probe coil is reduced to zero or near zero.

CLPV:
automating said process by utilizing a computer and a data bank, said data bank comprising data specifically applicable to each said probe, each said probe having computer readable identification marking thereupon such that on insertion into said NMR system said computer may identify each said probe and choose, from said data bank, appropriate data comprised of a recipe for demagnetizing said superconducting coils in each said probe; and

CLPV:
supplying the correct current to said demagnetizing coils such that the magnetization of each said superconducting probe coil is reduced to zero or near zero.

CLPV:
automating said process by utilizing a computer and a data bank, said data bank comprising data specifically applicable to each said probe, each said probe having computer readable identification marking thereupon such that on insertion into said NMR system said computer may identify each said probe and choose, from said data bank, appropriate data comprised of a recipe for demagnetizing said superconducting coils in each said probe;

CLPV:
supplying the correct current to said demagnetizing coils such that the magnetization of each said superconducting probe coil is reduced to zero or near zero.

ORPL:
Withers, et al., entitled "HTS Receiver Coils for Magnetic -Resonance Instruments", published in SPIE, Apr. 1994, vol. 2156, pp. 27-35.